THE EFFECT OF PRESSURE ON SOFT TISSUES

A REPORT OF A WORKSHOP SPONSORED BY THE
COMMITTEE ON PROSTHETICS RESEARCH AND DEVELOPMENT
DIVISION OF ENGINEERING — NATIONAL RESEARCH COUNCIL
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1972

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COMMITTEE ON PROSTHETICS RESEARCH AND DEVELOPMENT
of the
DIVISION OF ENGINEERING
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COMMITTEE ON PROSTHETICS RESEARCH AND DEVELOPMENT
DIVISION OF ENGINEERING-NATIONAL RESEARCH COUNCIL
NATIONAL ACADEMY OF SCIENCES--NATIONAL ACADEMY OF ENGINEERING

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PREFACE

For a number of years the Committee on Prosthetics Research and Development (CPRD) has come to recognize that the results of therapeutic treatment in individuals with neuromusculoskeletal disorders are affected very largely by the effects of force, and pressure, on living tissue.

There are very few procedures carried out by orthopaedic surgeons that do not involve the application of pressure to human tissue. Certainly, the very objective of every orthosis is the application of pressure to the human body, whether its primary purpose is to support, assist, or resist. The application of artificial limbs imposes pressure on soft tissues. The problems of decubitus ulcers, brought about by excessive pressure, are well known to all who are involved in the care of patients with spinal-cord injuries. Yet, very little is known about the effect of pressure on human tissues.

One of the reasons little is known about the effect of pressure on human tissues is that measurement of the pressures imposed is a very difficult problem. Therefore, to provide a background for a conference on the effect of pressure on human tissues, the Committee sponsored in May of 1968 a Workshop on Pressure and Force Measurement. At that time the various methods of measurement were reviewed and discussed, and the results were made available widely.

Responsibility for a conference on The Effect of Pressure on Human Tissues was given to the chairman of the Subcommittee on Fundamental Studies, CPRD, Victor Frankel, who called upon Donald Kettelkamp to head up a steering committee for carrying out this objective.

The steering committee* met at the Public Health Service Hospital, Carville, Louisiana, on January 12, 1971. At this meeting it was agreed that the subject of effect of pressure on human tissues was too broad for one conference; indeed, the subject of the effect of pressure on soft tissues was too broad to be treated during any one conference.

Accordingly, it was recommended that a series of workshops be held on the subject, the first being restricted to the effect of pressure on soft tissues.

The presentations in this report follow the outline of the original schedule (Appendix A). It is regretted that Richard Herman was unable to be present to discuss "Problems in Management of Other Neurological Diseases," that Edward Peizer was not able to report on work that had been carried out on measuring pressures involved in the immediate postsurgical fitting program, and that this report does not include the paper presented by Fred Caldwell.

A list of participants is given in Appendix B.

CPRD is most appreciative of the assistance received from the staff of the Texas Institute for Rehabilitation and Research in conducting this workshop. Especial thanks are due Mrs. Velda Montgomery, secretary to William Spencer, Director, TIRR.

* Donald Kettelkamp, *Chairman:* Colin McLaurin, Joseph Traub, George Lambert, Paul Brand, Frank W. Clippinger, Jr., Herbert Elftman, Richard Herman, Vert Mooney, and A. B. Wilson, Jr.

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SECTION I. STATEMENT OF OBJECTIVES

PROBLEMS IN MANAGEMENT OF AMPUTEES AND PARALYTICS

Vert Mooney

The tissue problems of amputees are quite a bit different from those of paralytics, and therefore require different management procedures. The amputee usually has adequate sensation, except in the case of some diabetics, so that the normal defense mechanisms against trauma are present. It is apparent, however, especially in the diabetic, that mild, chronic trauma may be tolerated for prolonged periods of time without much awareness on the part of the patient, and in time may result in considerable tissue breakdown before the patient becomes concerned. The two conditions that must be avoided are excessive shear and extremes of either positive or negative pressure. In the suction socket of the above-knee amputee, negative pressure frequently causes tissue breakdown when the pressure is prolonged and excessive. It is seldom that constant positive pressure causes ulceration; more frequently ulcerations are the result of minor repetitive impacts and tangential, or shear, forces. Of significance, of course, is the force-relieving ability of the soft tissues between the support device (prosthesis) and the load-bearing device of the body (bone). When there is a minimum of subcutaneous tissue present, all the problems indicated above are accentuated.

Thus, the solution to meeting problems caused by these clinically destructive forces lies in finding an interface that can substitute adequately for deficient subcutaneous tissue yet provide an environment where a minimum of shear forces will occur (adequate suspension), and the swings between negative and positive pressure are avoided, thus avoiding in turn the crisis of impact. Total-contact fitting techniques are aimed at solving the pressure swing problem; various types of liners such as those of Plastazote, the air-cushion socket, etc., bring about a better distribution of forces, but we still have great difficulty in providing adequate suspension. This is essentially the greatest technical problem facing improved prosthetic fit at present.

Turning to the problems of the paralytic, we still have the destructive elements of tangential and direct loads being transferred from an external support to the rigid, bony structure of the skeleton. However, no longer do we have sensation as a defense mechanism to provide a warning against excessive forces. We are talking about devices such as wheelchair cushions and mattresses for nonambulatory patients.

We are now aware that there are limits of pressure which cannot be exceeded for prolonged periods of time before tissue breakdown occurs. The amount of pressure that can be tolerated is related directly to the duration of time it is applied, but the maximum safe pressure that can be tolerated for prolonged periods in the buttocks area is 50 millimeters of mercury (arteriolar pressure). To a certain extent, the severity of the problem can be alleviated by training the patient to relieve himself of pressure from time to time, and by designing devices that can do this automatically. Nonetheless, the limits of safety cannot be exceeded. Unfortunately, pressure sores on the buttocks and backs of patients with spinal-cord injury are still extremely common.

In addition to distribution of pressure, there are practical considerations in providing the seat cushion or mattress that is useful in everyday activities. If the seat cushion is too heavy, the patient cannot transfer it. If it contains fluid or air, leakages will surely develop. Moreover, the simplistic approach of "floating-on-a-fluid" does not reduce the loading below 50 millimeters of mercury if the specific gravity of the fluid is close to unity.

One solution to the problem of seat cushions is to mold a cushion to the firm profile of the anatomic contours so that total, equal support can be provided without reliance on surface tension of the containing membrane. Evaluation and measurement of the patient are extremely critical in this approach in that the anatomy about the buttocks is extremely variable and tissue response varies widely among patients. Methods of measurement and prescription of support devices are critical to the successful use of pressure-relief equipment. In addition, recognition of factors other than direct pressure as a source of tissue breakdown is important. Thus, body positioning, local tissue environment, and activity levels of the patient are extremely important and must be evaluated carefully. Here again, anticipating the degree to which the destructive forces are due to tangential (shear) forces is relevant. We need methods of evaluating the variations in tissue tolerance to these forces that are more accurate than our current "looking-for-the-red-mark" method.

In summary, we need to understand more about support materials; we need better methods of measuring the forces which have the potential for destruction; and we need to know more about the variations in human response to these forces. Finally, we need better-organized clinical experience so that improvements and variations in design can be recorded accurately.

PROBLEMS RELATED TO INSENSITIVE LIMBS

Paul Brand

It is convenient to consider separately the problems of patients who are (a) insensitive and who are also severely paralyzed, such as paraplegics, and (b) insensitive but who are physically active, such as leprosy patients, diabetics, nerve-injured patients, etc.

(a) Paraplegics have difficulty in shifting their weight frequently enough so that circulation can be restored to the soft tissues which are subjected to pressure. They are unlikely to move forcefully enough to bruise or blister their tissues by the direct effects of loading. Their problems are mostly secondary to ischemia. The forces that may harm them range mainly from 0.5 PSI to 5 or 10 PSI, but these forces have to be sustained for an hour or more to begin to do real damage.

(b) Patients with diabetes, leprosy, and other peripheral sensory neuropathies, however, usually have no difficulty in moving around. They rarely stay still long enough to maintain ischemia (more than an hour). Their trophic ulcers usually come from repetitive loading of short duration and of levels of 30 to 300 PSI and more. However, problems can develop from continuous pressure while wearing a shoe or socket that is too tight.

Our investigations have been mainly with this second type of patient and with the levels of force that can cause direct damage to soft tissues.

Because insensitive limbs tend to suffer ulceration and absorption even though prolonged ischemia is avoided, the following questions demand an answer:

1. Is ulceration and absorption a direct result of denervation? We already know that the answer to this is "no." The damage occurs only in patients who are relatively strong and active. If a limb is protected from external force, it does not become absorbed or ulcerated.

2. If external loading is necessary to cause damage, does it take less force or loading to damage insensitive tissues than normal ones? We do

not know what the thresholds for damage are in the normal or in denervated tissue. We have demonstrated that tissues will accept very high forces without damage

a. if the force is applied only once for a short time. The same force or a lesser force applied repetitively over a longer time may result in bruising or other tissue damage;

b. if the force is applied evenly, as the pressure of water on a submerged object. Where shearing and distortional forces are applied, damage occurs at a lower level of loading; and

c. if the tissue is suitable for the acceptance of external loading. The tissue complex of the sole of the foot and the palm of the hand are apparently highly suitable for acceptance of repetitive loading. The dorsum of the hand is less so but has a capacity for absorption of tangential forces. Scar tissue may have tensile strength but too little compliancy to modify and moderate shearing and distortional forces.

Perhaps the most important function of the sensory nerves in the protection of the tissues is to sense the lowering of the threshold to damage that accompanies the repetition of the load. The threshold of pain becomes lowered and forces the subject to alter his grip or his gait until fresh areas of soft tissue get to bear the load while the previously loaded areas get to recover. The patient without sensation is able at the beginning of the day to accept repetitive loading as well as a normal person. However, as the repetitive loading continues and the tissues approach the threshold at which they will be damaged, the patient has no way of knowing that this is occurring so he doesn't change his gait or his grip and goes on until he is bruised or blistered.

In order to be able to help such patients in the fitting of shoes and prostheses and in the organization of their activity to prevent soft-tissue damage, studies need to be organized with the following objectives:

To design and make available a transducer to measure pressure. Such a transducer must be thin (less than a millimeter), rugged, and inexpensive. Its ready availability to every orthotist and prosthetist is more important than its absolute accuracy.

A transducer that will measure tangential and horizontal shear forces.

A system for measuring the distribution of force over the whole surface of an extremity.

We need to study the thresholds for damage of normal and denervated tissues, recognizing that each tissue probably has a different threshold and that tissue complexes may be more or less efficient at resisting stress than single isolated tissues, and recognizing, too, the factor of time and repetition.

Because of the complexity of such measurements, we should also develop a system for determining the response of human tissue to stress in order to make the prosthetist, as well as the patient, aware when tissues are being subjected to stresses greater than they should be allowed to tolerate. It may be possible to recognize this by changes in the tissues at a stage when we are still unable to produce instrumentation to analyze all of the forces to which the tissue is being subjected.

SECTION II. BASIC BIOLOGIC CHARACTERISTICS OF TISSUES AND SYSTEMS

HISTOCHEMISTRY OF SKIN--AN ABSTRACT

Jeffrey S. Pinto

The skin is uniquely designed to resist distortion by various natural forces. The cutaneous structures responsible for this resilience seem to reside in the dermis. The dermal collagen, by virtue of its high tensile strength, opposes stretch and shear forces, while the dermal proteoglycans, in concert with the dermal collagen network, resist compression by virtue of their strongly hydrophilic nature and resistance to displacement.

Under normal conditions the skin is exposed to a myriad of "subthreshold" shear, stretch, and pressure forces every day. The skin reacts to these forces in a predictable manner and is not damaged by them. However, if that threshold is exceeded the skin is damaged to a degree dependent on the extent and duration of the force.

Shear forces result in slight dermal inflammation, and in either blister or abrasion of the epidermis, depending on the strength and thickness of the stratum corneum, the strength of the dermo-epidermal junction, and the degree of binding of the skin to subcutaneous structures. Small stretch forces take up the "slack" in the dermal collagen network. Once this slack is taken up, further tension results in "slippage" as the collagen becomes extended; under extreme tension, the tensile strength of the collagen is exceeded, and the skin ruptures. Mild pressure of short duration produces mild ischemia followed by reactive hyperemia and does not result in permanent damage. Pressure of greater magnitude or longer duration can result in inflammation, hemorrhage, ischemia, pressure necrosis, bacterial infection, and ulceration. It seems, however, that the skin is much more resistant to the damaging effects of pressure than other tissues which have a smaller proteoglycan content.

THE MICROCIRCULATION

Mary P. Wiedeman

It is the purpose of this presentation to define the microcirculation and describe its components and their functions.

The term "microcirculation" is used to designate blood flow through small vessels. The microcirculatory bed is that portion of the vascular system concerned with the transfer of gases and nutrients and is frequently referred to as the exchange system. These vessels should be considered separately from the vessels of the macrocirculation which is comprised of the larger vessels designed for the maintenance of systemic blood pressure.

The final ramifications of the distributing vessels of the arterial system represent the beginning of the terminal vascular beds of the microcirculatory system. Vessels which are recognized as components of the microcirculation are arterioles, terminal arterioles, capillaries, postcapillary venules, and venules.

The microcirculatory system begins with arterioles. Branches from arterioles are terminal arterioles, so called because they terminate in a capillary network and have no vascular communication, such as an anastomosis or arcuate structure, with any other arterial or venous vessel. The precapillary sphincter is the name given to the final smooth muscle cell that separates arterial and capillary vessels, and it is that portion of the terminal arteriole that contracts or relaxes to monitor blood flow into the capillary network that lies distal to it. The capillary need only be defined as the pure endothelial tube that lies between distributing arterial vessels and collecting venous vessels. The proximal and distal limits are not easily defined by direct microscopic observation, although it may be said to begin where vascular smooth muscle disappears and to end where convergence of two vessels marks the beginning of the collecting venous system. In some vascular beds, the vessel which results from convergence of two capillaries is called a postcapillary venule.

Arterial and venous vessels accompany one another except in the final ramification of the arterial vessels in the capillary network. Arterial vessels are smaller in diameter than their venous companions and have thicker walls. There are many more venous vessels to collect blood and return it to the heart than there are distributing arterial vessels.

It should be made clear that the exact diameter of a vessel is not the basis on which a vessel is placed in any designated category, because the dimensions of a vessel will vary considerably depending on the organ, tissue, or species of mammal being documented. It is more important to consider the location (and therefore function) of a vessel in the vascular distribution, at the same time recognizing that there is great similarity in vascular architecture among all mammals at the level of the terminal beds.

Consideration should be given here to the frequent speculation that anatomical shunts exist between arterial and venous vessels, thereby bypassing the capillary network. The apparent need for such an arrangement is to establish a means by which blood can be moved with great rapidity from the arterial to the venous vessels. The idea that such structures exist is substantiated to some extent by investigations, but anatomical shunts have not been liberally demonstrated by *in vivo* microscopic techniques. Except for numerous arteriovenous anastomoses in the rabbit ear and dog's paw, both of which are used in temperature regulation of these animals, and in some parts of the G.I. system, the evidence that there are such anatomical structures is still controversial. They occur so infrequently in most vascular beds that have been seen in living animals that it is hard to believe that they can have a truly significant influence on total flow from arterial to venous vessels. On the other hand, if a careful study is made of the structure of terminal vascular beds, it is apparent that blood can traverse the area between the arterial and venous systems rapidly and effectively and therefore satisfy the conditions for which shunts have been postulated.

Frequently, in the same bed, one sees a short pathway, a pathway of intermediate length, and a long, tortuous route between the arterioles and the venules. It is reasonable to assume that metabolic demands of the tissue and hemodynamic forces subserving the total organism would dictate whether all, some, or only one route would be open at any one time. Because these pathways are outside the macrocirculation and therefore not involved in pressure maintenance, the systemic blood pressure need not be compromised by variations in the flow through these networks.

One of the most outstanding features of the behavior of both arterial and venous vessels is their spontaneous contractile activity which is called vasomotion. Vasomotion is generally seen in arterial vessels as a slowly developing contraction which reduces the internal diameter of the vessel to varying degrees and may occur at the point of branching of an arterial vessel, along its entire length, or in segments of the vessel. The contraction may last for a considerable period of time and then slowly diminish, or it may be rapid in the onset and cessation.

Characteristically the venous vessels are involved in a spontaneous rhythmical contraction, the rate of which can be shown to vary with the amount of blood contained in the vessel at that time. Both venous vasomotion and some arterial vasomotion are independent of nervous control. The activity continues in the denervated venous vessels with little modification and is enhanced in denervated arteriolar vessels that were not under control of the sympathetic nervous system to begin with.

The vascular architecture and their activities described here were shown as recorded on color movie film through the microscope.

[Dr. Wiedeman closed her presentation by showing a most interesting film.]

THE EFFECT OF MECHANICAL STRESS ON DENSE CONNECTIVE TISSUE

John W. Madden

Chronic mechanical stress has an obvious and profound effect on dense connective tissues. Predicting the order and magnitude of these effects, however, is difficult. For example, prolonged mild stress on periarticular structures produces hypermobile joints in dancers and athletes; in contrast, prolonged stresses applied to joint structures altered by immobility may only increase stiffness. In wounds subjected to chronic stress, hypertrophic scars may develop; in wounds without tension, scars are usually fine and delicate.

Although the outcome of mechanical stress is difficult to predict, in each instance stress seems to influence the form and physical properties of the dense connective tissue. Because the physical characteristics of all mesenchymal tissues depend on the physical properties of collagen fibers and the collagen molecules they contain, the metabolism of collagen may provide a clue as to how these changes occur. Connective tissues become stronger by (1) increasing the total density of collagen; (2) increasing the density of intermolecular covalent cross-bonds within the fibers; or (3) altering the architecture of the collagen fiber network. In the healing wound, changes in physical properties seem to occur by the third mechanism. In spite of a stable collagen content, healing incised wounds demonstrate a rapid collagen turnover for many months. Scanning electron micrographs confirm the slowly changing architectural patterns of scar collagen. This dynamic metabolism of collagen demonstrated in the healing wound seems to be present in most dense connective tissues. Mechanical stress may affect shape and physical properties by influencing the turnover of collagen. Currently, experiments are being conducted to measure directly the effect of stress on the kinetics of collagen metabolism.

SECTION III. SUMMARY OF WORK CARRIED OUT TO DATE

THE CLEARANCE RATES OF RADIOSODIUM FROM SUBCUTANEOUS TISSUES SUBJECTED TO EXTERNAL PRESSURE*

James B. Reswick

Radiosodium tracer techniques were used to develop an understanding of the way in which various body fluids and metabolites move through tissue spaces under the influence of external pressure. It is assumed that the small radiosodium (Na^{24}) ions injected into the subcutaneous tissues will diffuse through the interstitial fluid spaces and into the capillaries in the same manner as the sodium normally present. The radiosodium, after crossing the capillary wall, will be carried swiftly away in the blood stream. The changes in the count rate measured by a scintillation detector can therefore be considered as an index of blood flow through the region. When external pressure is exerted over the tissue it causes an impairment of the normal blood flow which can be assessed by measuring the degree of reduction of the radiosodium outflow rate.

Human subjects were tested while resting either in an armchair or supine in bed. One and one half microcuries of Na^{24} in 0.05 ml of normal saline solution were injected subcutaneously, through a No. 26 gauge needle, into the volar surface of the middle forearm which was supported horizontally in the supine position. Pressure levels of between 40 and 300 mm Hg were exerted over a circle 1.25 in. in diameter for desired periods of time by the head of the scintillation detector resting over the site of radiosodium injection. The desired level of pressure was obtained either by application of a weight or by means of a pneumatic piston and cylinder. For the "zero pressure" test, i.e., to measure the normal outflow rate, the detector was so counterbalanced that it just touched the skin.

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* Excerpted from the paper on "Deformation and Flow in Compressed Skin Tissues" by K. E. Hickman, O. Lindan, J. B. Reswick, and R. H. Scanlan from Western Reserve University Medical School and Case Institute of Technology. Published in Biomedical Fluid Mechanics Symposium, Denver, Colorado, April 25-27, 1966, pp. 127-147, by The American Society of Mechanical Engineers, New York.

Adult male albino rats were held snugly in a cylindrical container which had a longitudinal slit over the back through which a skin fold was pulled. The skin fold was suspended vertically by two surgical threads piercing the edges. The radiosodium was injected subcutaneously into the middle of the suspended skin fold and the desired pressure levels between 20 and 100 mm Hg were obtained by means of a pneumatic piston and cylinder, the piston pressing upon the skin fold held against the scintillation detector. The circular pressure area was 0.75 in. in diameter.

In the tests on humans the scintillation detector crystal was separated from the skin only by a thin protective metal plate. In the experiments on rats, the crystal was situated within a hollow lead tube 1.5 in. away from the animal skin.

As the radiosodium moving away from the injection site was distributed within the whole body, it continued to contribute to the count rate recorded by the scintillation detector over the pressure area. In humans no correction was made in the count rate recorded over the pressure area as the number of counts recorded over the other uninjected forearm was only slightly above the background.

In the rat experiments, however, the radiosodium present in the vicinity of the pressure area and in the rat's body contributed markedly to the counts recorded by the scintillation detector placed over the pressure area. It was determined that approximately one third of the counts originating from the radiosodium outside the pressure area was recorded by the scintillation detector at the end of the experiments, and the data had to be corrected accordingly. At the end of each experiment the animal was sacrificed, the skin flap with the pressure area was separated from the animal, and the animal carcass was removed, while the pressure area was left compressed at 200-300 mm Hg against the scintillation detector for counting. The difference in the counts before and after the carcass was removed enabled the determination of the counts originating from the rat's whole body.

In order to provide a convenient measure of the clearance rate of radiosodium from tissue subjected to external pressure, the rate of decrease in counts was expressed as an outflow half-time. The outflow half-time (OT) was defined to be the time in minutes required for the number of counts per minute to drop to half of its initial value.

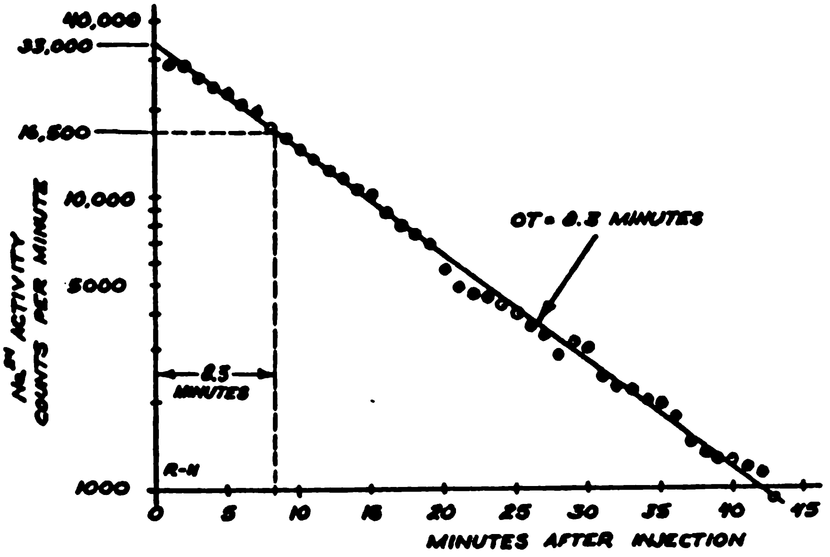

Fig. 1. Normal clearance of radiosodium from the human forearm. No external pressure was applied. It took 8.3 minutes for the number of counts to be halved, e.g., from 33,000 cpm to 16,500 cpm.

Immediately after the subcutaneous injection of radiosodium, the normal outflow half-time was determined without any external disturbance to the tissue. Then the pressure was applied and the resulting outflow half-time was determined. The change in the radiosodium outflow for the compressed tissue was expressed as the ratio

$$\frac{(OT_{\text{pressure}})}{(OT_{\text{no pressure}})}$$

High values of this ratio indicated marked reduction in the radiosodium outflow from the compressed area.

RESULTS OF RADIOSODIUM TRACER STUDIES

EFFECT OF PRESSURE LEVEL

In man, the effect of various levels of externally applied pressure on reduction of outflow rates of radiosodium from the subcutaneous tissues of the forearm is illustrated in Figure 2. At 40 and 60 mm Hg the outflow rates were slowed down to 60 and 40 per cent, respectively, of their normal value. At pressure levels of 80, 100, and 150 mm Hg, outflow rates were only 13, 7, and 4 per cent of their original value.

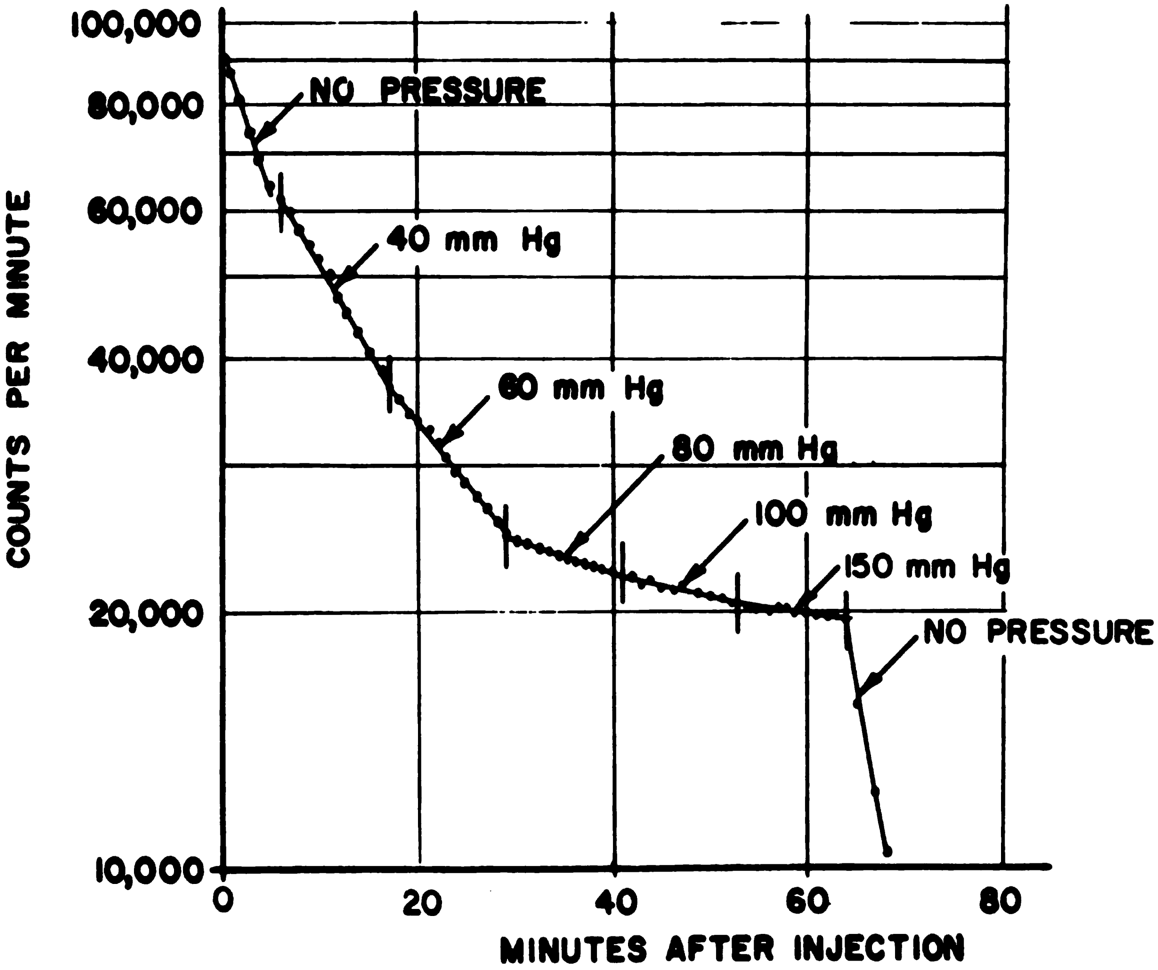

Fig. 2. The clearance rates of Na^{24} from the forearm subjected to increasing levels of external pressure.

It is accepted that the blood pressure within human capillaries is about 25 mm Hg, and even lower in the venules. However, application of this level of pressure to the skin of the forearm did not completely close the subcutaneous capillaries and venules, the clearance rate of radiosodium being still 89 per cent of the normal. Fifty mm Hg were required to reduce the outflow of sodium by half and 75 mm Hg to reduce it to 25 per cent.

When similar tests were carried out on the human forearm and on the rat's back skin fold, the results shown in Figure 3 were obtained. The pressure level which reduced the outflow of radiosodium to 10 per cent of its normal value was 75 mm Hg for rats and 95 mm Hg for humans. The greater impairment of circulation due to the effect of external pressure in the rat's skin when compared with that of the human forearm may be explained, in part, by a different manner of pressure application. The compression of the rat skin fold between two hard surfaces assured a more uniform pressure distribution within the area of tissue examined.

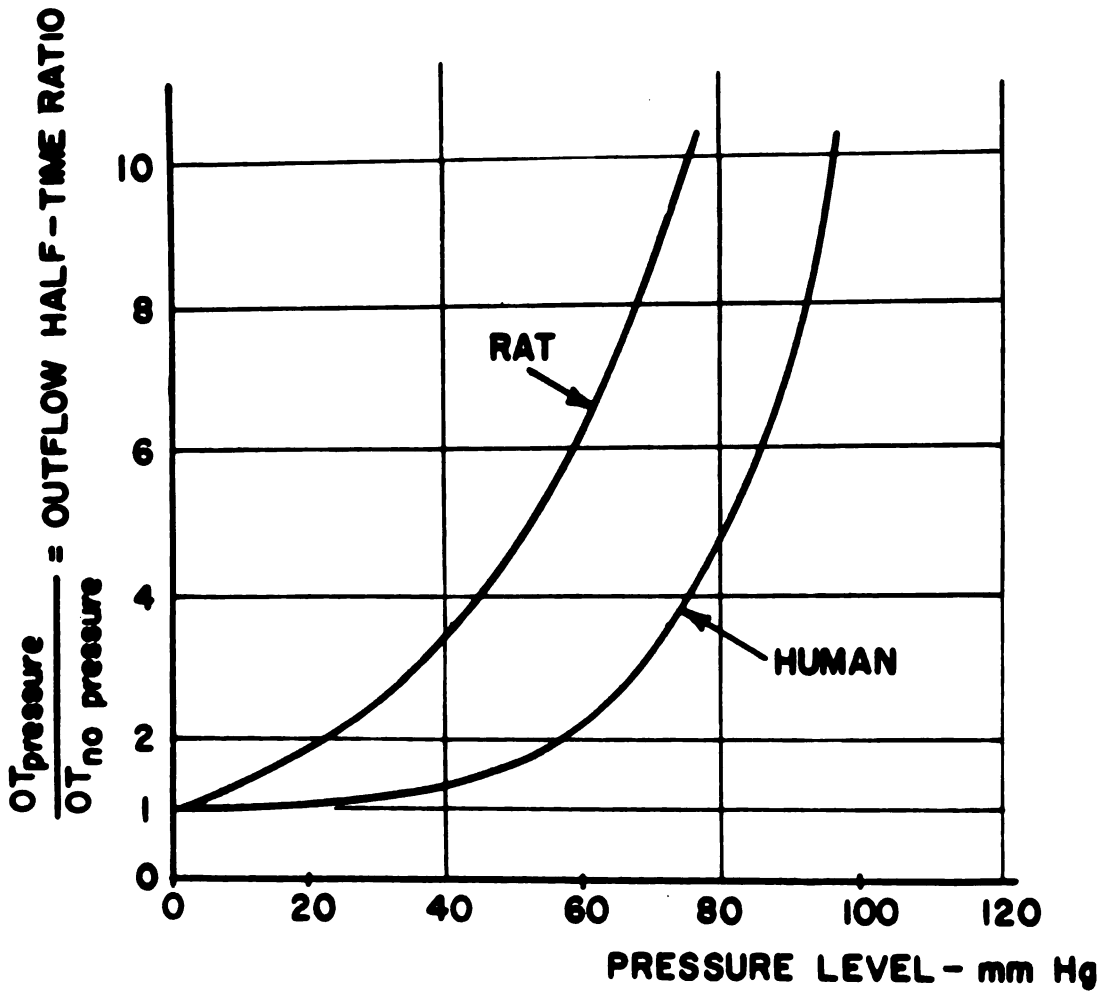

Fig. 3. Effect of external pressure on the outflow half-time of radiosodium from subcutaneous tissues of rat and man. Curves are mean values of several tests.

EFFECT OF EXPOSURE TO PROLONGED PRESSURE

When a low level of pressure (e.g., 40 mm Hg) was applied over the radiosodium injection site on the human forearm, the sodium outflow rate decreased with time. A series of curves is shown in Figure 4. It was found that the initial rate of outflow continued more or less unchanged for 10 or 15 minutes, thereafter gradually decreasing by a factor of 1.5 to 4.2 in a series of experiments.

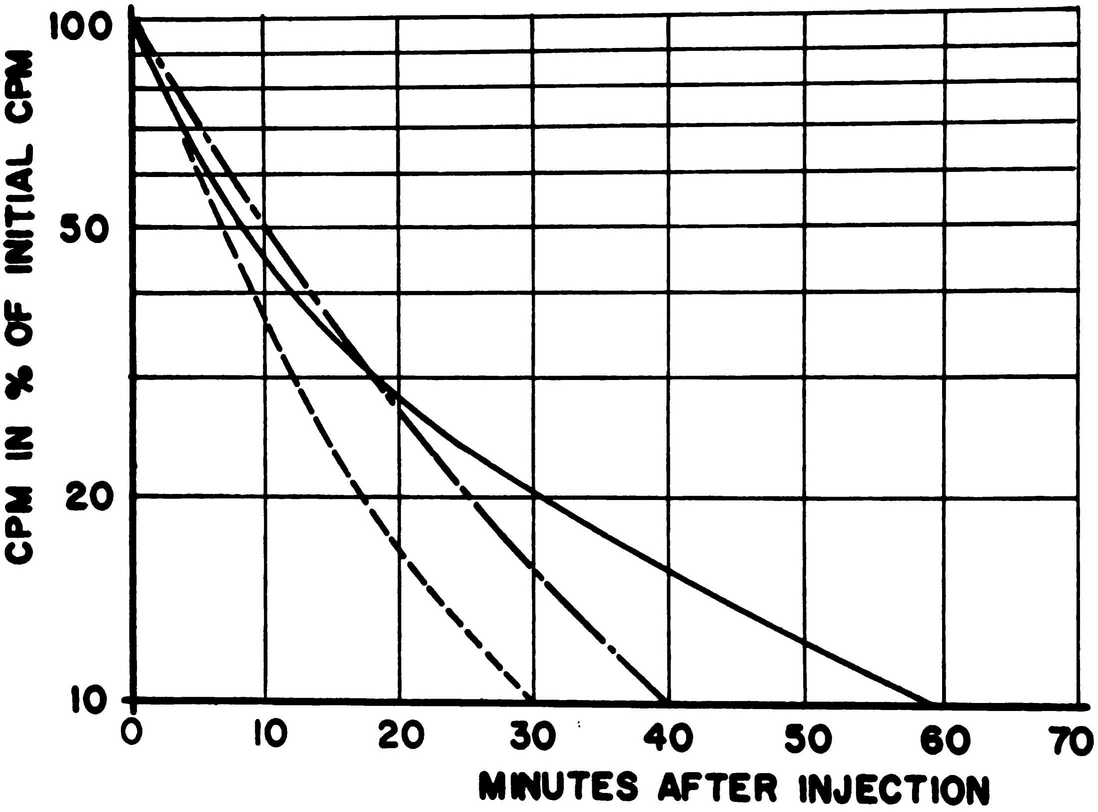

Fig. 4. Gradual reduction of outflow rates of radiosodium from the human forearm subjected to local pressure of 40 mm Hg for 30 to 60 minutes.

EFFECT OF GRADUAL RELIEF OF PRESSURE

It was found that when the level of external pressure was gradually increased and then gradually decreased, the outflow rates of the subcutaneous radiosodium followed different patterns as shown in Figure 5. The total experimental time from zero to 110 mm Hg and back was one hour. It followed that the relationship between the external pressure level and the

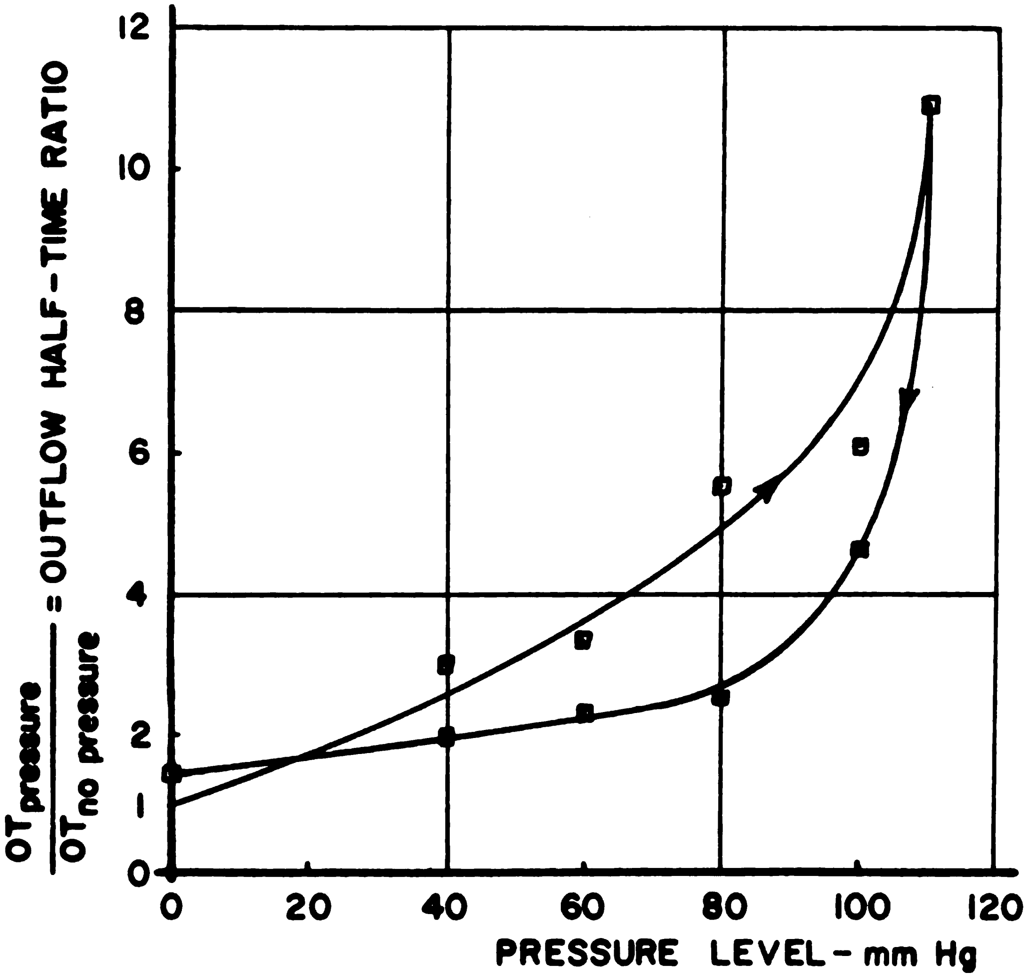

Fig. 5. Effect of rising vs. falling levels of pressure on outflow of radiosodium from the human forearm. The arrows indicate rising or falling pressure levels.

peripheral circulation was not straightforward but depended on previous exposure to pressure.

CORRELATION OF RADIOSODIUM OUTFLOW WITH TISSUE RHEOLOGICAL BEHAVIOR

The results of the human forearm tests were useful in revealing the relationship of pressure level and duration to the impairment of circulation. However, it was not possible to measure the concomitant rheological and radiotracer testing. The engineering "boundary conditions" were better controlled in these skin fold experiments and simultaneous recordings of sodium clearance rate, pressure level, and tissue deformation could be obtained. Typical results are shown in Figure 6.

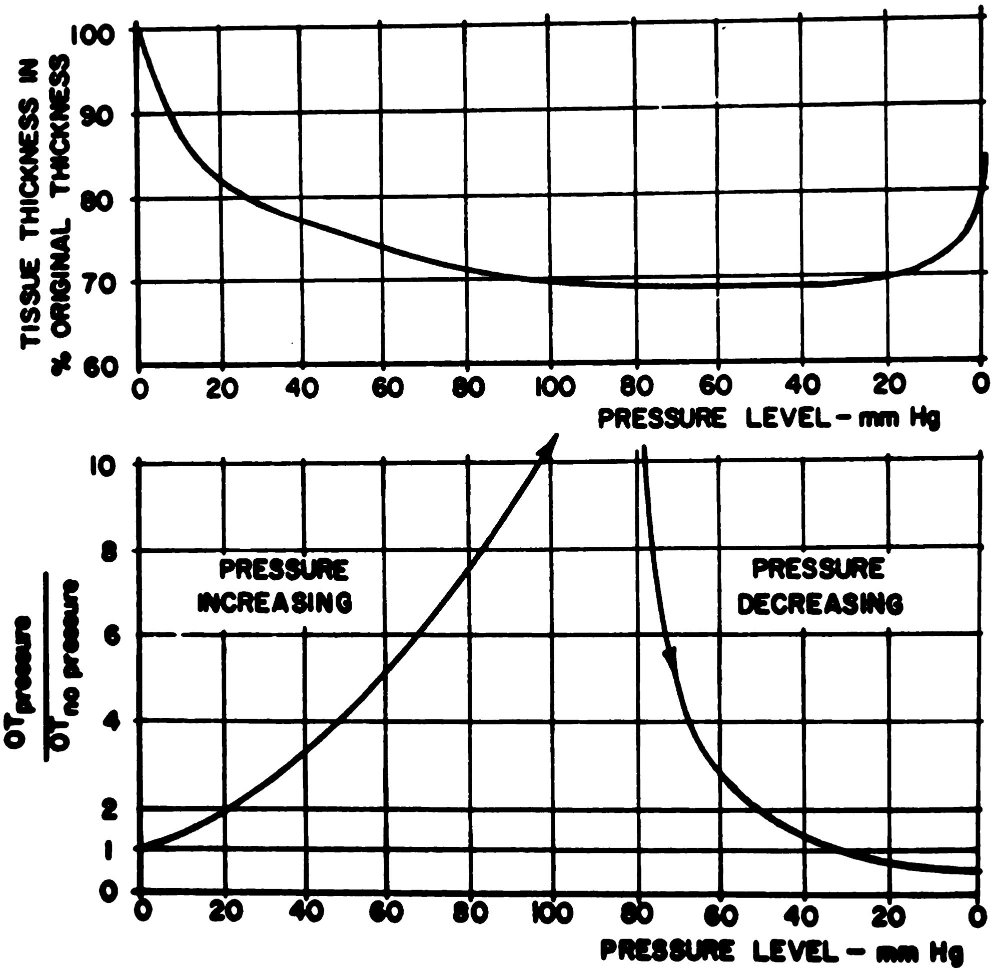

Fig. 6. Relationship between tissue compression and concomitant reduction of radiosodium outflow.

The simultaneous measurements of the radiosodium outflow rate showed very clearly the early return of circulation when pressure was gradually released, although very little recovery in tissue thickness was taking place. Even more surprising was the finding that at about 30 mm Hg of pressure, when the tissue thickness had hardly begun to recover from compression, its circulation was becoming better than the normal flow as estimated before application of pressure.

Although some allowance should be made for the fact that the skin was stretched in the experimental setup, the fact remains that the circulatory recovery preceded the viscoelastic recovery from the effects of external pressure.

SUMMARY AND CONCLUSIONS

The effect of external pressure on the flow of blood and metabolites within the soft tissues was studied using radiosodium tracer techniques. Special instruments and test procedures were developed.

It was found that levels of external pressure well above capillary pressure could be supported by skin and subcutaneous tissues before blood flow was seriously impaired. When a constant, moderate level of pressure was maintained, the clearance rate of sodium dropped slowly as time passed. When the level of pressure was gradually increased up to a selected maximum, and then gradually reduced back to zero, the circulation returned at much higher pressure levels than would be expected from the increasing-pressure tests. When the deformation of the tissues was measured simultaneously with the radiosodium clearance rate, it was found that the return of the circulation even preceded any appreciable change in deformation of the tissue.

From these test results, a preliminary and qualitative "engineering model" can be proposed to describe the rheological behavior of skin and subcutaneous tissue subjected to external pressure. This model will be described in terms of the tissue response to the application of a prolonged, constant level of pressure and its subsequent relief--i.e., to the creep-test procedure.

Immediately upon application of pressure, the tissue deforms through an elastic, accordion-like compression of the connective tissue matrix in the derma and subcutaneous layers of the skin. Cellular matter is displaced primarily by "rigid-body" motion, and some flow of the less-viscous fluids in the tissue occurs. Thus, the external pressure loading is resisted by elastic forces in the connective tissue matrix, compressive and shear stresses in the soft cellular matter, and the elastic "membrane" forces of the skin itself.

While the constant pressure is maintained, the tissue continues to deform gradually in a manner similar to a Voigt material. This viscoelastic compression may be due to flow of the viscous cellular matter caused by internal pressure and shear stress gradients established during the immediate compression. As a result of this viscous flow, some of the external pressure loading is transferred from the elastic elements of the tissue to the tissue fluids and to the capillaries, thus causing a gradual reduction in blood flow. This reduction is superimposed upon the larger reduction in blood flow caused by the immediate compression of the tissue when pressure was first applied.

When the creep-test pressure level is increased, the deformation increases, but not in direct proportion. As the tissue becomes more and more deformed, its resistance to further deformation increases, as might be expected for such a porous, fluid-filled elastic matrix. Circulation is maintained when the external pressure is well above the capillary pressure because much of the pressure load is carried by the connective tissue matrix and the cellular structure surrounding the capillaries and larger vessels. It is only at high-pressure levels that the deformations become great enough to block the capillaries.

When external pressure is completely removed, the tissue recovery contains both elastic and viscoelastic components. These responses probably take place by mechanisms which are the reverse of those postulated for immediate and delayed compression. However, some of the components of the tissue return only very slowly to their original shape and position. This delay in recovery causes a "plastic" molding deformation to remain after delayed elastic recovery is essentially completed.

If the external pressure is removed gradually, the return of circulation precedes any appreciable change in deformation of the tissue. If it is assumed that the amount of pressure which can be carried by the elastic components of the tissue is a constant at any particular level of deformation, then the difference between this elastic resistance and the external pressure must be made up by compressive stresses in the softer tissues. These stresses tend to block capillary flow. As the external pressure is reduced, the difference between the load and the elastic resistance falls, thus lowering the compressive stresses in the softer tissue and permitting the capillary flow to resume. These postulates are consistent with the concept of tissue as a porous elastic connective-tissue matrix filled with viscous cellular matter and liquids.

FUTURE WORK

It is recognized that the engineering model proposed above for the rheological behavior of skin and subcutaneous tissues has not been thoroughly confirmed by the test programs reported. The engineering boundary conditions of the skin and subcutaneous tissue tests could not be controlled carefully. Furthermore, accurate assessment of the anatomical and chemical changes in tissue composition, concomitant with the level and the duration of external pressure, will be required.

ACKNOWLEDGMENT

The technical assistance of R. W. Corell, C. R. Glasener, T. W. Dennison, H. D. Huge, W. L. Schultz, and Mrs. Elizabeth Scanlan is gratefully acknowledged.

This investigation was supported in part by Research Grants No. RD-695 and RRTC-5 from the Vocational Rehabilitation Administration of the U.S. Department of Health, Education, and Welfare, and also by Special Fellowship No. 1-F3-GM-20, 536-01 awarded by the National Institute of General Medical Sciences, Public Health Service, U.S. Department of Health, Education, and Welfare.

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INFECTION AS A CONTRIBUTORY FACTOR IN THE PRODUCTION OF EXPERIMENTAL PRESSURE ULCERS

Rosemary Lindan, O. Lindan, and W. K. Massalski

Bacterial infection is invariably present in pressure ulcers. The possible etiological role of the infecting organisms in the production of such ulcers has not been determined. The technical ability to produce a graded series of standardized "sterile" lesions on the skin of healthy animals by pressure alone[1] afforded an opportunity for observing what, if any, contributing effect local, and systemic, bacterial infections might have on the development of experimental pressure ulcers.

STANDARD LOCAL INFECTIVE LESION

The first step in the experiment was to define and establish a "standard local infective lesion" to be used in conjunction with standardized pressure lesions. The desirable "standard" local infection was defined as being: reproducible, limited to erythema and edema, not progressing to necrosis of superficial tissues, producing no serious generalized reaction, and resolving without sequelae within a given period. Such a lesion was consistently produced in healthy rabbits by the following procedure:

> Staph. pyogenes, var. aureus, bacteriophage type 80/81, recently isolated from a patient who died from an infection with this organism, was grown for 18 hr. at 37°C in Tryptic Soy Broth (Difco). The broth culture was spun down and the organism resuspended in sterile saline before use. The infecting dose was 0.01 ml of the bacterial suspension injected as an intradermal bleb into the ear of the rabbit on the dorsal surface. The infecting dose was found by plate counts to contain 10^7 colony-forming particles.
>
> The resulting lesion was an initial bleb, which disappeared within approximately 30 min. and was followed by an area of erythema and edema, 6 - 7 mm in diameter, which became maximal in 24 - 48 hr. It subsequently became a small nodule, decreasing gradually in size and color, and disappearing completely by the 7th day.

THE EFFECT OF CONTINUOUS EXTERNAL PRESSURE ON THE COURSE OF A LOCAL INFECTION

METHODS

Using the techniques described above, the rabbit ears were injected with staph. pyogenes and two hours later pressure clips were applied over the sites of injection and kept in place for 7 hours at a continuous pressure of 100 mm Hg (right ears). Controls (left ears) were of two types: a) with "standard local infections" only; b) with pressure clips applied for the same length of time over the sites of injection of 0.01 ml of sterile saline.

RESULTS

It was found that pressure of this degree and duration prevented the early localization and walling off of the infection. Figures 1, 2, and 3 demonstrate that after 24 hours the ears subjected to both pressure and infection showed a diffuse and marked erythema and edema, extending beyond the confines of the pressure area. Localization of the infection was not appreciable until 72 hr. after the application of pressure and frequently proceeded to frank ulceration. Figures 4 and 5 show the appearance of experimental ears 96 hours after application of injury.

Fig. 1. Appearance of rabbit's ear 24 hours after application of injury. Injury consisted of 7 hours of continuous pressure of 100 mm Hg without infection.

Fig. 2. Appearance of rabbit's ear 24 hours after application of injury. Injury consisted of standard local infection without pressure.

Fig. 3. Appearance of rabbit's ear 24 hours after application of injury. Injury consisted of standard local infection followed by 7 hours of continuous pressure of 100 mm Hg.

Fig. 4. Appearance of rabbit's ear 96 hours after application of injury. Injury consisted of standard local infection without pressure.

Fig. 5. Appearance of rabbit's ear 96 hours after application of injury. Injury consisted of standard local infection followed by 7 hours of continuous pressure at 100 mm Hg.

EXACERBATION OF LOCAL INFECTIVE LESION BY THE APPLICATION OF INTERMITTENT EXTERNAL PRESSURE

Because the application of continuous pressure for 7 hours resulted in a delay in the localization of the infection, an experiment was designed to see whether intermittent pressure for a longer period (which had been found to cause no permanent tissue injury under "sterile" conditions) would entirely prevent localization of the infection and result in a systemic bacterial invasion.

METHODS

For this series of experiments staph. pyogenes 'phage type 52 A/79 was used because it was found that in systemic infections with the 80/81 strain the rabbits frequently died rapidly from toxemia. The local lesions produced by the 52 A/79 strain were identical, macroscopically, with those produced by the use of the 80/81 strain. Intermittent pressure of 100 mm Hg was applied for a total of 14 hours, with five-minute-relief periods every hour, over the site of the bacterial infection.

RESULTS

Neither the infective lesion by itself nor the pressure injury alone resulted in gross tissue necrosis. The combination of the two, however, produced a local abscess and ulceration within 24 hours. No bacteremia was detected by daily blood culture nor, on sacrificing the animals, was any pathological evidence found of systemic infection.

FACTORS CONCERNED IN THE PRODUCTION OF A LOCAL INFECTED LESION AT THE SITE OF PRESSURE IN BACTEREMIC RABBITS

METHODS

Rabbits were injected intravenously with 10^7 - 10^8 live staphylococci of the 52 A/79 strain to produce a chronic low-grade bacteremia. We had confirmed previous findings of Rogers[4] that this strain disappeared from the bloodstream shortly after injection and reappeared 48 to 72 hours later and continued to be detectable for several days subsequently.

In the first series of experiments 100 mm Hg pressure was repeatedly applied on the second, third, and fourth days following the i.v. injection of organisms. In each instance the pressure was sustained for seven hours at the same location on the ear.

RESULTS

The level of bacteremia produced was found to be critical in determining the local tissue response to pressure. With bacteremia of 10^3 orgs./ml or more the animal became very ill, and showed signs of bacteremic shock, i.e., the ears were cold and very pale, cyanosis was evident around the mouth and nose, and the animals were listless and breathing rapidly. In these animals the compressed tissue showed little inflammatory reaction, in fact much less than that of the uninfected controls. On postmortem examination macroscopic abscesses were found in the internal organs but no macro- or microscopic abscesses were found at the site of pressure on the ears.

With bacteremias of less than 10^3 orgs./ml the animals showed only the standard response to pressure of this degree and again no gross abscess formation was detectable on the ear or elsewhere in the body.

In the second series of experiments local pressure injury was induced prior to the intravenous injection of the organisms. The right ears of the rabbits were subjected to 14 hours of continuous pressure, and the left ears to 7 hours of continuous pressure. Twenty-four hours later the rabbits were injected intravenously with approximately 10^7 viable staphylococci. The animals were sacrificed on the ninth experimental day. Histological examination of the ears showed pus formation at the site of the 14-hour-pressure lesions only, and culture from these areas yielded the 52 A/79 strain of staphylococci. Sections of the left ears subjected to 7 hours of pressure showed no abscess formation and little residual inflammatory reaction by the ninth day.

COMMENTS ON ROLE OF INFECTION IN DECUBITUS FORMATION

LOCAL INFECTION AND PRESSURE

There can be no question but that the effects of local infection and local pressure reinforce each other to a marked degree. When pressure was applied continuously over the freshly infected site so that there was no access of blood or leukocytes to the area, the subsequent localization of the infection was delayed and the area of inflammation extended well beyond the pressure area.

Of particular interest was the demonstration that the application of 14 hours of intermittent pressure with 5 minutes' release every hour over the site of infection produced most intensive necrosis. Evidently the alternation of hyperemia with ischemia predisposes most to decubitus formation in the presence of infection.

BACTEREMIA AND PRESSURE

The rabbits in bacteremic shock demonstrated clearly the inability of their tissues to respond with an inflammatory reaction to a local noxious agent.

In the animals with milder bacteremias the local pressure lesion had to be of a severe nature before there was evidence that the blood-borne infection had localized at the site of pressure.

ACKNOWLEDGMENT

This investigation was supported in part by Research Grants No. RD-695 and RRTC-5 from the Vocational Rehabilitation Administration of the U.S. Department of Health, Education, and Welfare.

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RESULTS OF WORK AT PUBLIC HEALTH SERVICE HOSPITAL
CARVILLE, LOUISIANA

Paul Brand

TRANSDUCERS

We are no further ahead with the development of a pressure transducer than we were at the beginning of the study. The Filpip transducer which was available ten years ago is no longer in production. In our view, this transducer because of its relative small production cost is still the best transducer to use on a mass scale.

MICROCAPSULE FABRIC SYSTEM

A system has been developed for the assessment of the distribution of pressure over the surface of the limb. This has been done by impregnating microcapsules into a fabric which is then formed into a stump sock or slipper sock or glove. These microcapsules fracture under various pressures causing a staining on the sock which is proportional to the forces which impinge on that area. This system is still imperfect because it is difficult to calibrate in terms of actual psi or shearing force. However, it is at present by far the best system for giving a dynamic summation of all types of force impinging on tissue. It acts in a very similar way to pain threshold in a sensitive person because it responds more to shearing forces than it does to forces normal to the skin, and it responds more to repetitive loading than to a single load. The system is inexpensive and practical and could be made available to all limb fitters and shoemakers. It should be routinely used for fitting shoes to diabetics and other patients with lack of sensation. Work is continuing on this system in order to increase its shelf life.

REPETITIVE LOADING STUDIES

Studies of repetitive loading over a period of many months on the digits of monkeys seem to indicate that the sensitive digits and the insensitive digits respond in a similar way to the repetitive load. This suggests that the reason for the destruction of insensitive limbs is not a weakness of the tissues but is simply the fact that insensitive limbs tend to accept more force than sensitive ones do.

STUDIES CONCERNING DENERVATION

It has been observed that in the course of damage from denervation a progressive fibrosis takes place in the tissues. Instrumentation has been developed for measuring this change and for studying the viscous and elastic properties of soft tissue.

TEMPERATURE STUDIES

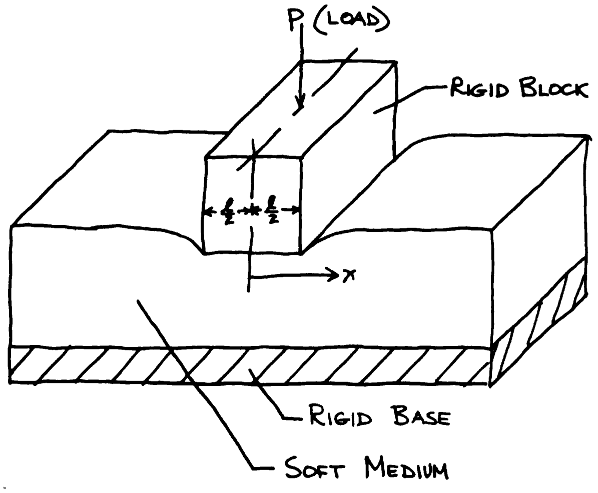
Recognizing that perhaps the best index of commencing tissue damage is commencing hyperemia, various methods of measuring the change of temperature of tissues are being studied. Currently, the Thermovision thermograph is proving to be a useful tool for identifying areas that have been subjected to excessive stress. The key to the use of temperature as an index of damage is that the actual temperature is less important than the temperature differential between the stressed area and a corresponding unstressed area, and also that an immediate rise of temperature is less significant than the continued maintenance of the temperature differential after time has been given for physiological adjustments and a return to normal.

MECHANICAL STRESSES WITHIN FLESH

Leon Bennett

Soft-tissue trauma is the result of imposing unacceptable loads upon flesh. What happens when a hard block (representing a brace or prosthetic socket) is pressed against flesh? Are the stresses produced within the flesh large--so large as to produce a failure analogous to "ultimate failure"; or are the stresses small, so that many loading cycles produce a failure analogous to "fatigue failure"? Do flesh stresses extend beyond the loading member, and if so is there a correlation between cyst location and stress as claimed by Allende, Levy and Barnes[1]? If we agree that stress is bad, can anything be done to reduce stress, perhaps by the novel design of loading members? These are the questions we are considering.

Our tools are theoretical. Theory is cheap and theory is powerful, but there is a catch. Theory is no better than the underlying assumptions. As an example, consider Sadowsky's[2] solution (Fig. 1) to the problem of a hard block pressing against a soft medium. If we let x approach $\ell/2$ the predicted pressure goes to infinity; which, of course, is not true. Here the failure of the analysis results from a postulated inability of the soft material to sustain large deflections and/or local plastic flow at the corners. This is a useful warning.



$$\text{local pressure within soft medium} = \frac{P}{\pi\sqrt{\left(\frac{\ell}{2}\right)^2 - x^2}}$$

Fig. 1. Sadowsky's solution.

In our model (Fig. 2), flesh is taken as a homogeneous mélange (points 1,2,3) of all its constituents aside from bone. Flesh deflection characteristics (points 4,5) are assumed known and fixed, although there is evidence that these vary with the magnitude of the applied load. Finally (point 6) we take flesh not to be viscoelastic, that is, all reactions are taken as independent of time.

FLESH IS CONSIDERED:

1. To include skin, blood, fat, muscle.
2. To exclude bone.
3. To be uniform.
4. To possess a fixed modulus of elasticity.
5. To possess a fixed Poisson's ratio.
6. NOT to be viscoelastic.

Fig. 2. Flesh model assumptions.

These assumptions represent great simplifications and help us get the job of flesh stress analysis done quickly--but are they warranted? We really don't know. Only experimental testing will decide. In the meantime, lacking experimental verification, it must be understood that our work may be in error.

We are concerned (Fig. 3) with the stress at a point M contained within a slab of flesh of thickness H, subject to loading q(x). Note that the loading is two dimensional, that is, every slice along the z axis is identical. Such a slice, of width δ (lower portion, Fig. 3), is labelled with the appropriate stresses, where σ_y represents the perpendicular stress to the upper face and σ_x the perpendicular stress to the vertical face. The shear stresses are given by τ_{xy} and τ_{yx}. So far we have computed solutions for σ_y on the basis that it appears to be the larger of the two normal stresses and therefore the more significant.

The derivation procedure (Fig. 4) uses the deflection approach, in which the principal unknown is the displacement v(x,y) of a point of interest M. For each load we can write the deflection contribution at M as a combined function of x and y dependent terms. This technique is known as

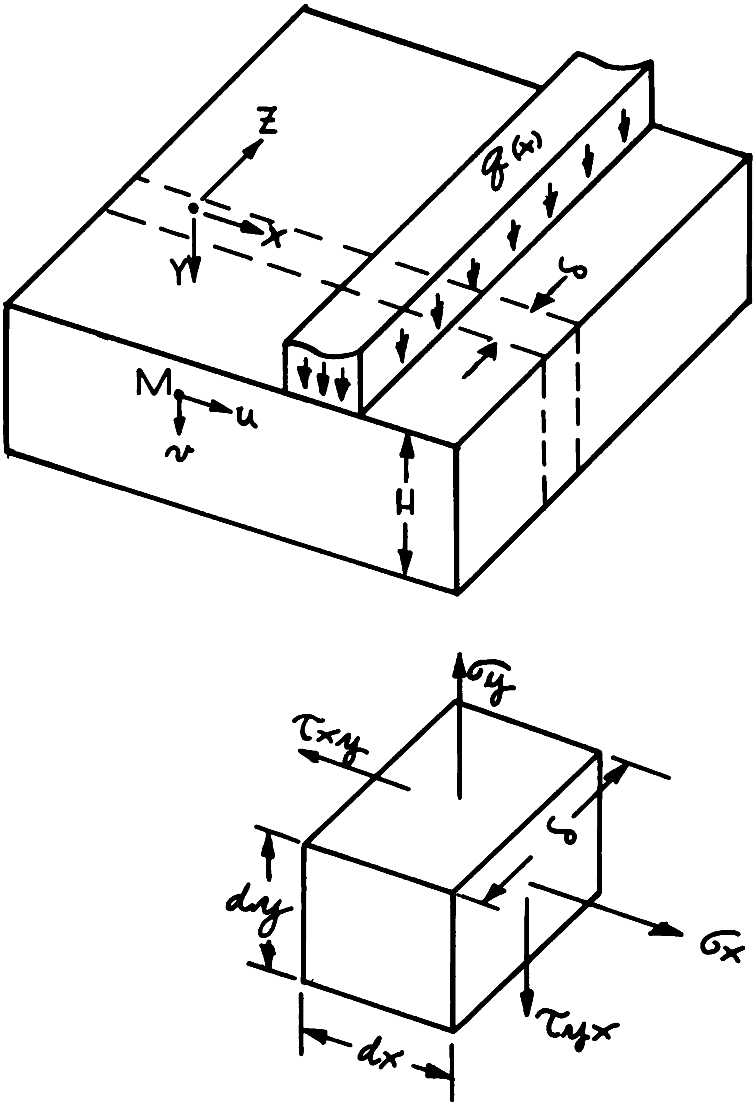


Fig. 3. Coordinate system employed.

the separation of variables. All loads and corresponding deflections are then summed up. The x dependent terms are solved by a combination of theory of elasticity techniques and strain energy methods. The y dependent terms have been obtained through empirical testing of a soft medium with limited plastic flow (soil). The full derivation is given elsewhere[3] and will not be repeated here. In general the derivation is well proven and becomes uncertain only in the use of the empirical term. It would be better to conduct appropriate tests in flesh. However, even if the empirical term is

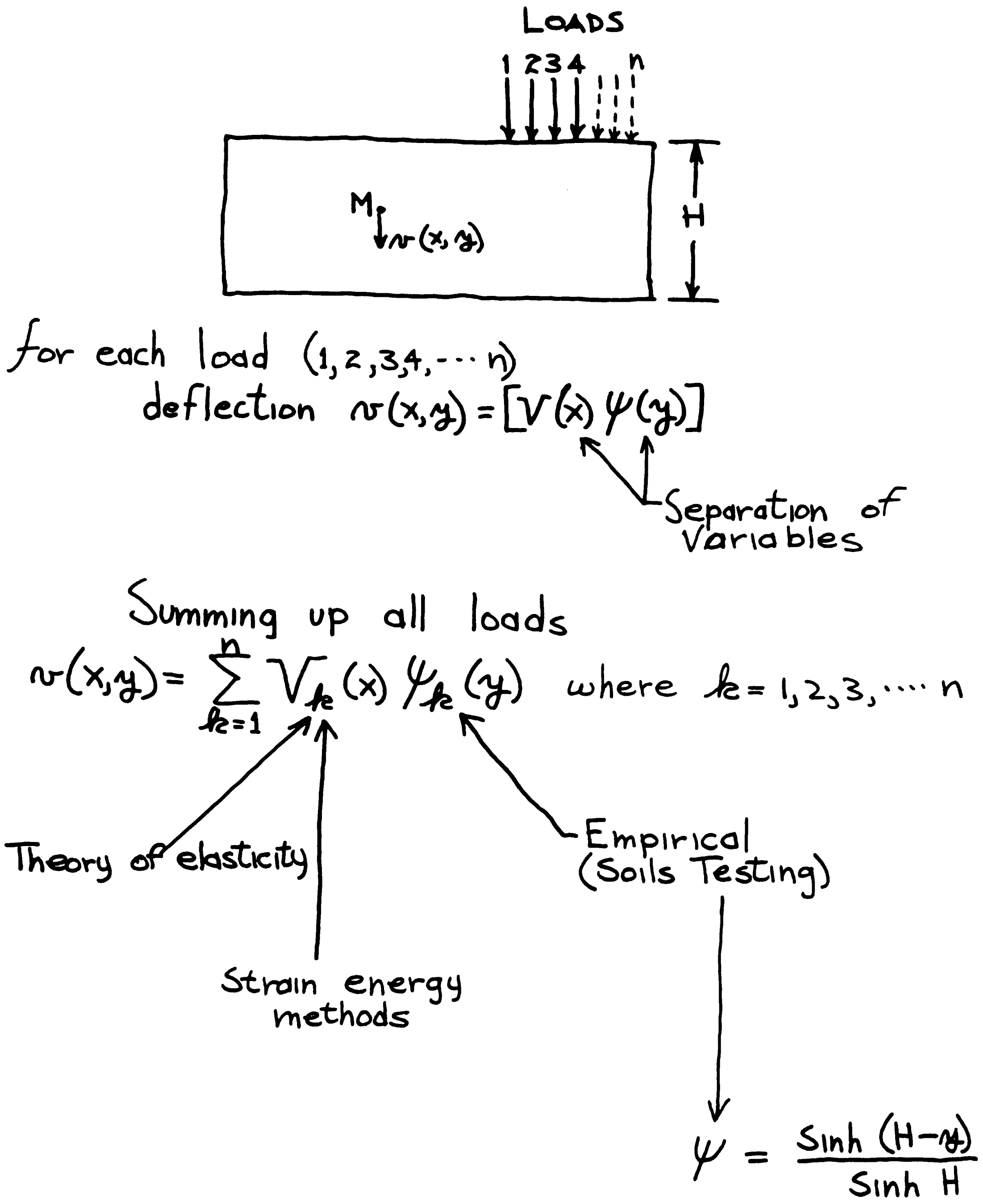


Fig. 4. Derivation procedure.

totally wrong, which is unlikely, only the vertical distribution of stress will be affected, and not the peak stress values. As we are looking primarily for large stress values, our major results will be unaffected even by a major error in the empirical term.

We have solutions for the following loading types (Fig. 5). In Case A we are pressing a dull chisel, or a wire, or a narrow brace against a thickness of flesh supported by rigid bone. In Case B, we have a similar situation to A, except that a thin sheet of plywood, say a tongue depressor, has been inserted between the loading edge and the flesh. In Case C the load is applied through a membrane or corset or bandage to the flesh. In Case D the load is applied through a wide block that is absolutely rigid. In each case the load is the same, P.

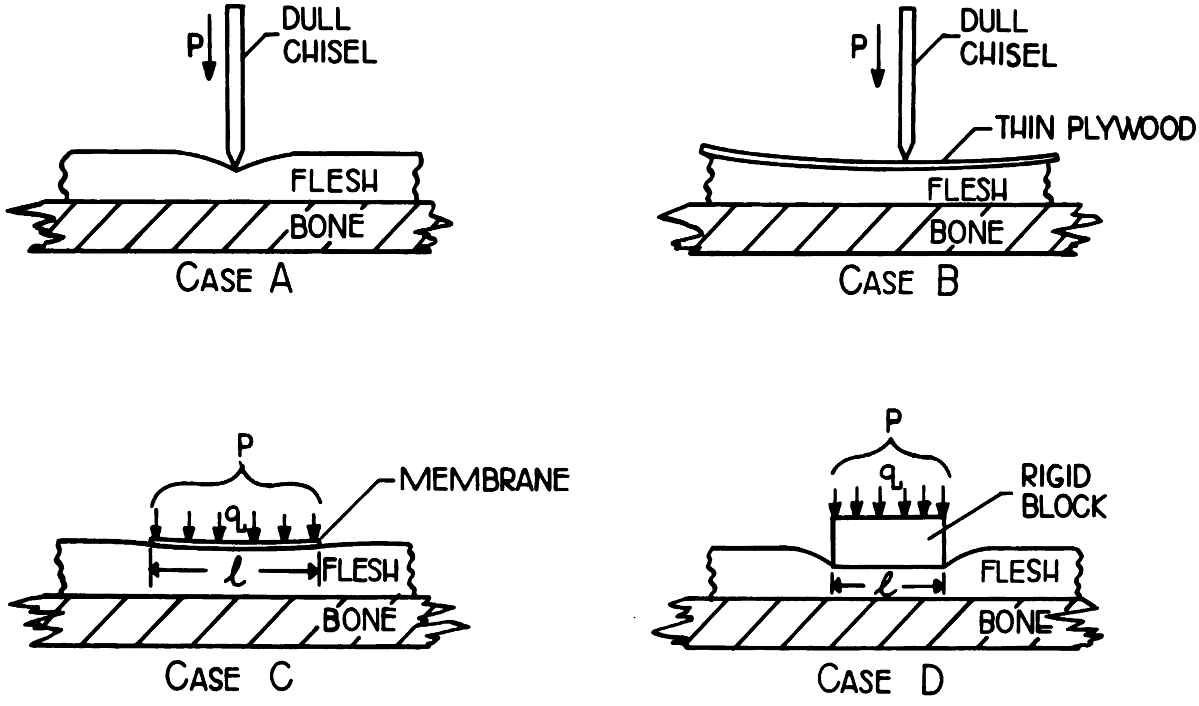


Fig. 5. Types of loading studied,

A numerical solution to Cases A and B is given in Figure 6. The numbers at a given location in the flesh cross section are the values of normal compressive stress per inch of z (i.e., $\delta = 1$) when P equals one pound, and the Poisson's ratio is taken as one third. If the load is say 10 lb., multiply the numerical answers by 10 to get the proper value. Stress units are standard (pounds per square inch). From left to right (Fig. 6) the cover thickness increases. From top to bottom the flesh thickness increases

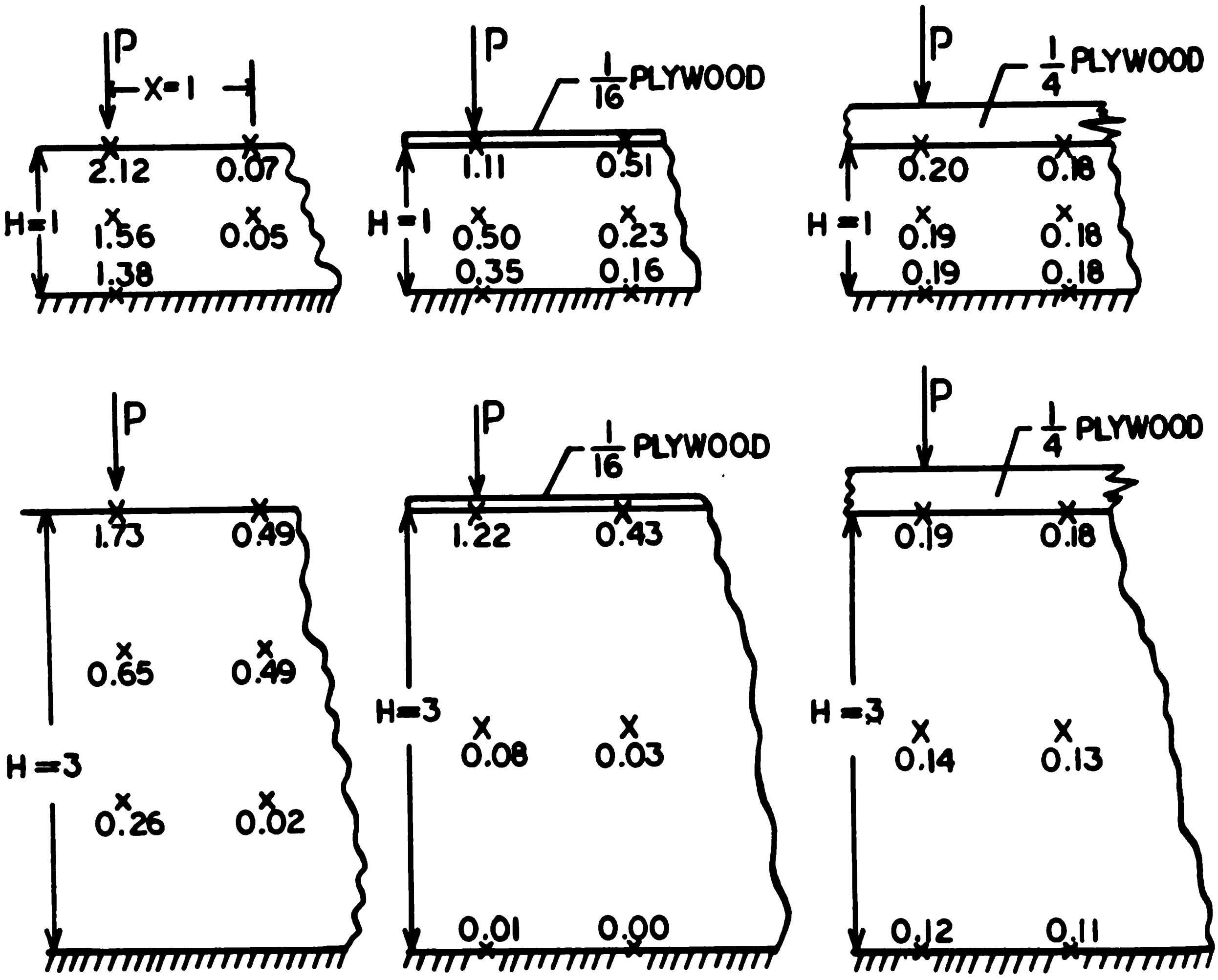


Fig. 6. Normal stress within flesh; cases A and B. Poisson's ratio flesh taken as 1/3. Stress at a locale equals numerical value shown times P.

(one inch to three inches). Note that even a thin cover cuts the maximum stress value for one inch flesh in half. As the cover thickness becomes large, we arrive at a low and nearly uniform distribution of stress throughout the flesh.

In the above computation we have used a Poisson's ratio* of one third. What limited information we have on flesh suggests that there is no fixed value of Poisson's ratio. Studies of cat skin indicate Poisson's ratio to be a function of the applied load; the higher the load the higher Poisson's ratio. It is of interest to examine the effect of Poisson's ratio variability (Fig. 7). As the Poisson's ratio increases, the effect is to

* The ratio of lateral contraction to longitudinal extension.

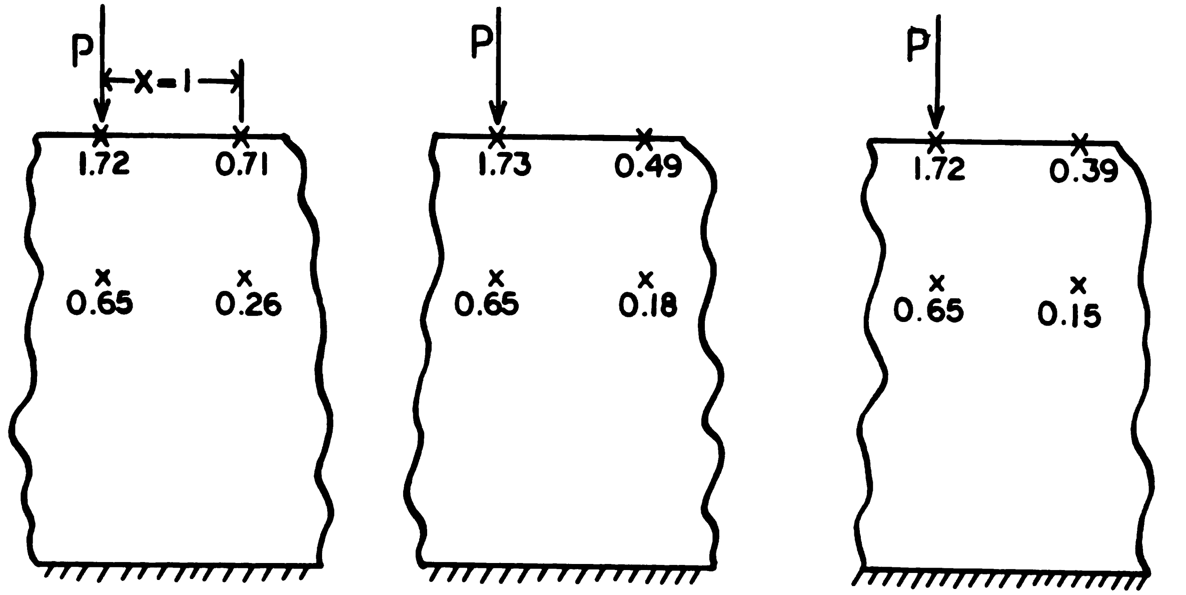


Fig. 7. Normal stress within flesh versus Poisson's ratio. Stress at a locale equals numerical value shown times P. Flesh thickness H=3. Case A load.

reduce stresses remote from the loading axis. Stress along the loading axis is not influenced. It would seem that under conditions of increasing load the results given here are too high at points remote from the loading site.

The remaining cases (Figs. 8 through 11) use a different means of presentation. The stress at a given level (surface or inner) is plotted as a function of the distance from the center line of the load. Case C (the uniformly loaded membrane) has been computed for membrane widths of 2, 6, and 20 inches and flesh thicknesses of 1 and 3 inches.

Examination of the thick flesh, Case C load (Fig. 8), shows that peak stresses are produced under the narrow width and the inner layer is always loaded to a lesser degree than the surface. Stresses beyond the edge fall off to zero in about an inch. Note that the stress level beyond the loading edge can be large.

Given the membrane load and a thinner section of flesh (Fig. 9), there is movement towards convergence of stress levels between surface and inner layers. As before the stress goes to zero in about an inch beyond the member.

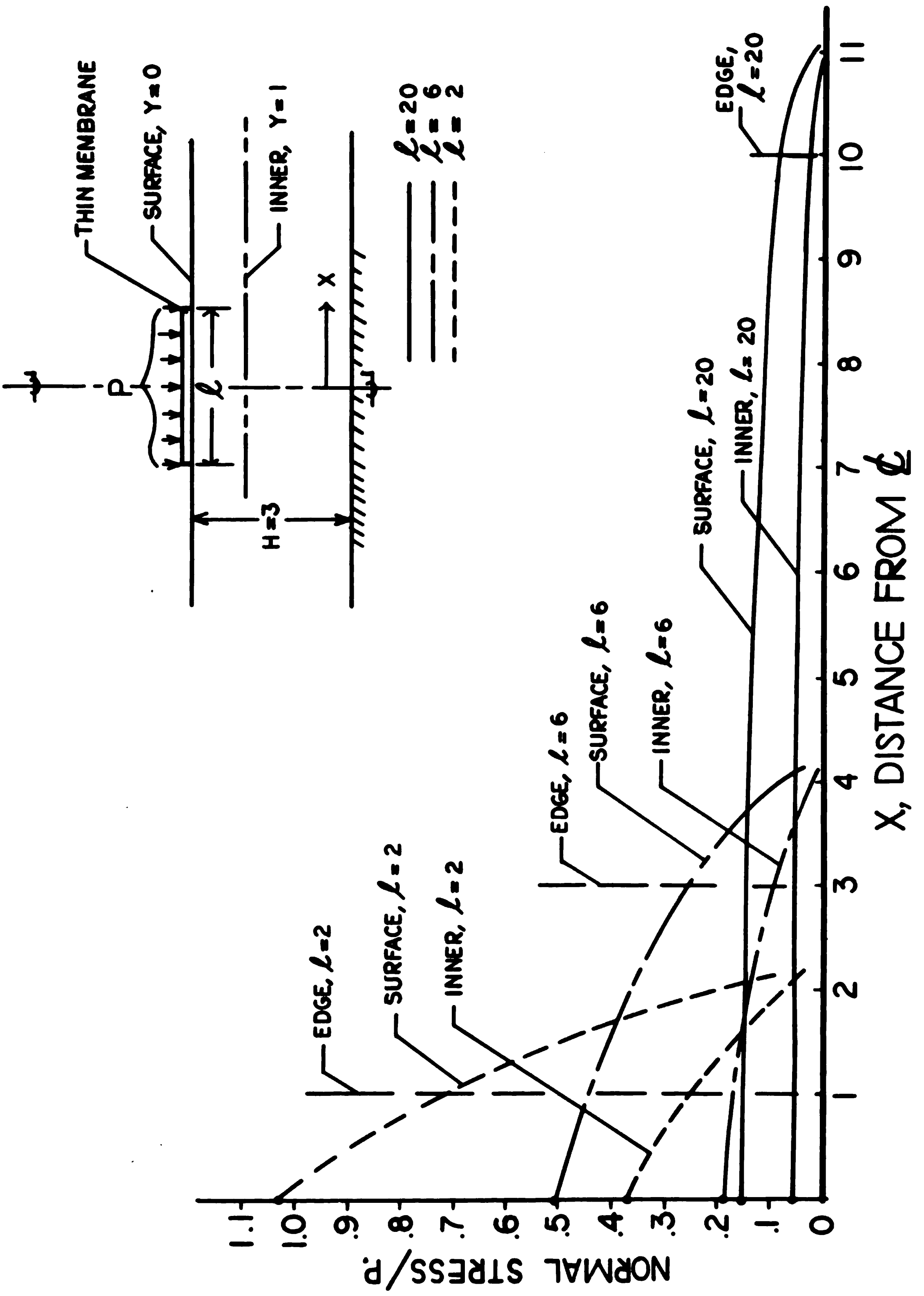


Fig. 8. Results, Case C, flesh thickness = 3

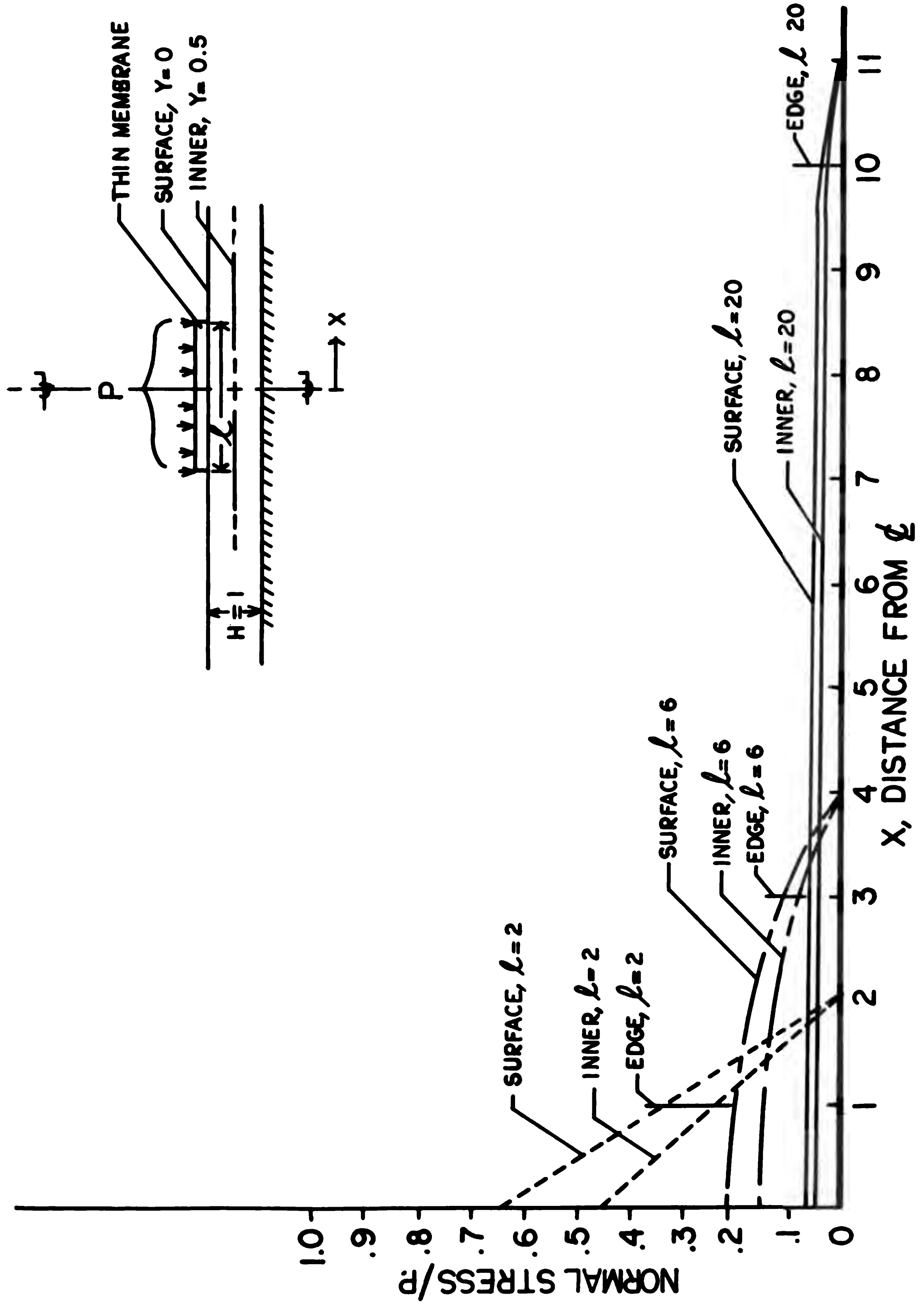


Fig. 9. Results, Case C, flesh thickness = 1

Given the rigid loading block and the thicker flesh case (Fig. 10) the stresses are flat beneath the block and then decay to zero in about an inch. Note that the magnitudes are lower than those under the membrane. The thinner flesh displays similar characteristics (Fig. 11), with the converging tendency even more strongly shown.

To sum it up: the dull chisel case (A) is the most severe, even a slender cover plate (B) greatly reduces the flesh stress. As the cover plate becomes rigid (D) the main effect is to even out the stresses. Stresses in that flesh beyond the direct action of the load decay to neglible values in about 1¼ inches.

So far we have dealt with compressive stress. We are now examining shear stress. In a rough, qualitative way, the shear stress is proportional to the gradient of compressure stress. For a typical compressive stress distribution (Fig. 12) shown as a solid line, the shear stress would resemble the dashed line. Note that the shear stress not only peaks beyond the load member edge, but it does not even exist under the load member. Such a distribution corresponds to the prediction of Allende and Levy[1] concerning cyst location. In their view, cysts are found largely beyond the brim of prosthetic sockets, and not inside the socket, because the shear is large outside the socket. It certainly appears that considerable shear exists outside the socket. This opens an interesting possibility, namely, the control of shear by controlling the compression stress gradient; in turn controlled by local socket stiffness. Conceivably such a process may result in the reduction or elimination of cysts and other skin trauma among the class of artificial-limb users.

Currently we are examining the shear stress levels for Cases A through D in a quantitative fashion.

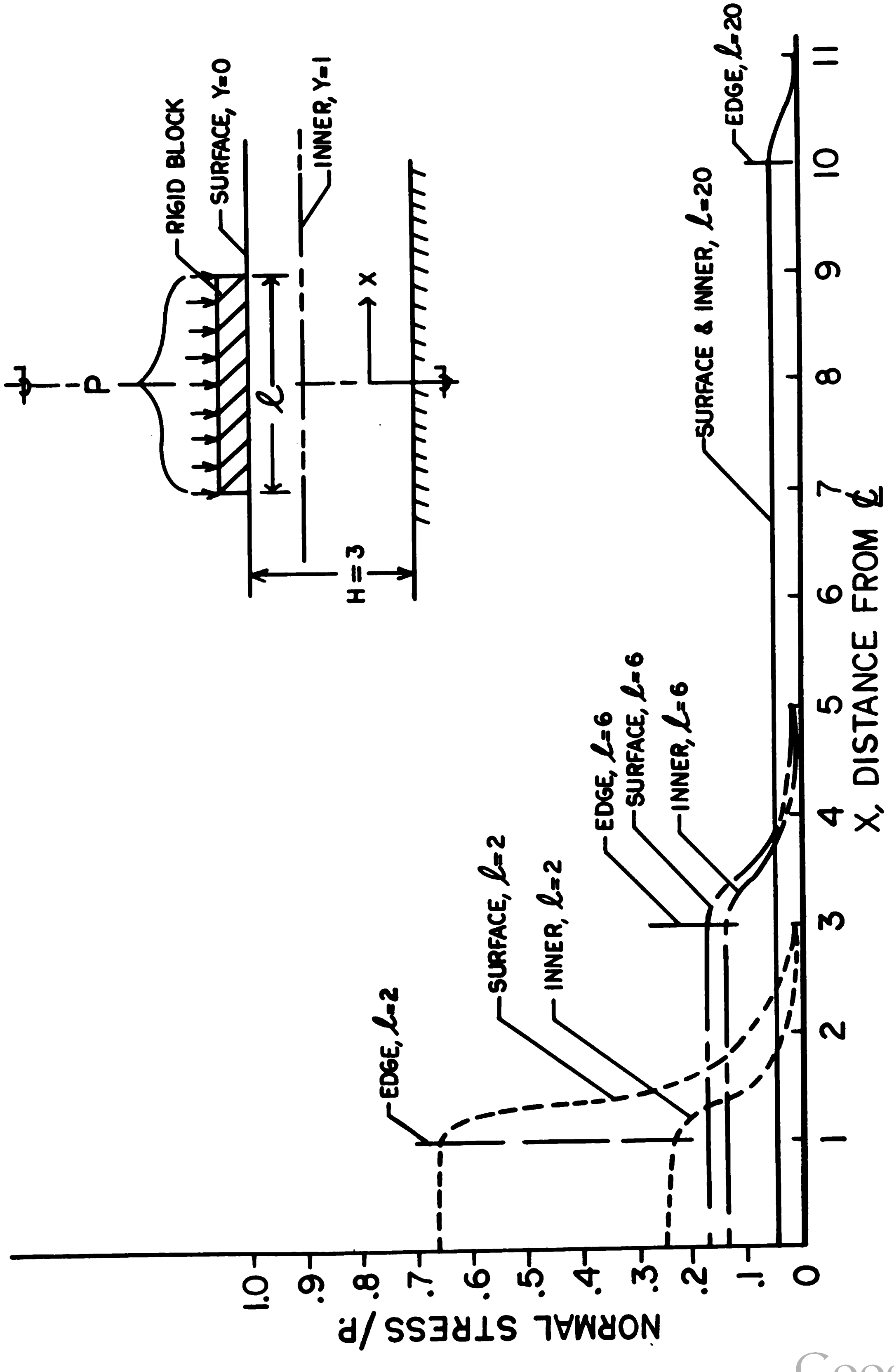


Fig. 10. Results, Case D, flesh thickness = 3

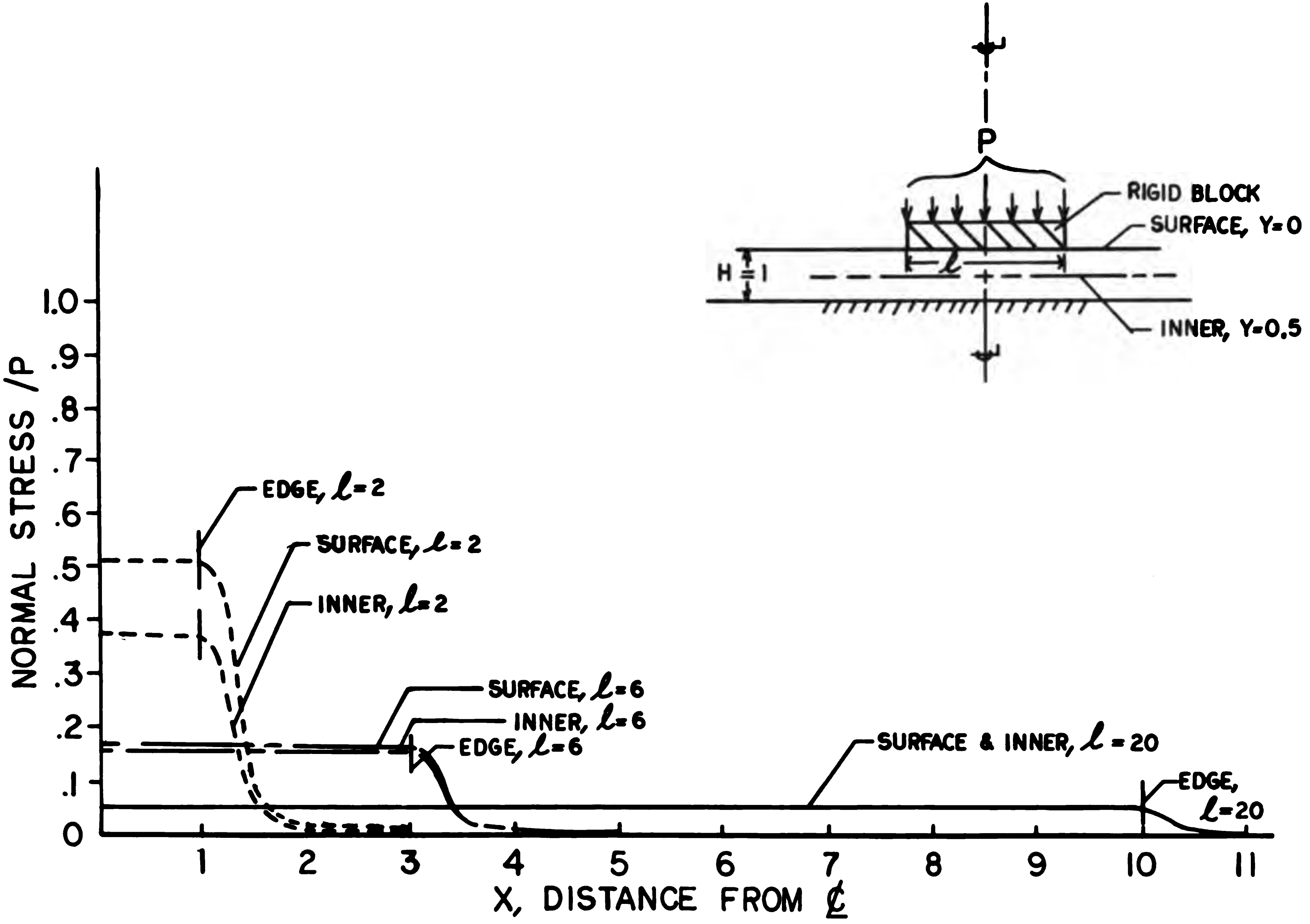


Fig. 11. Results, Case D, flesh thickness = 1

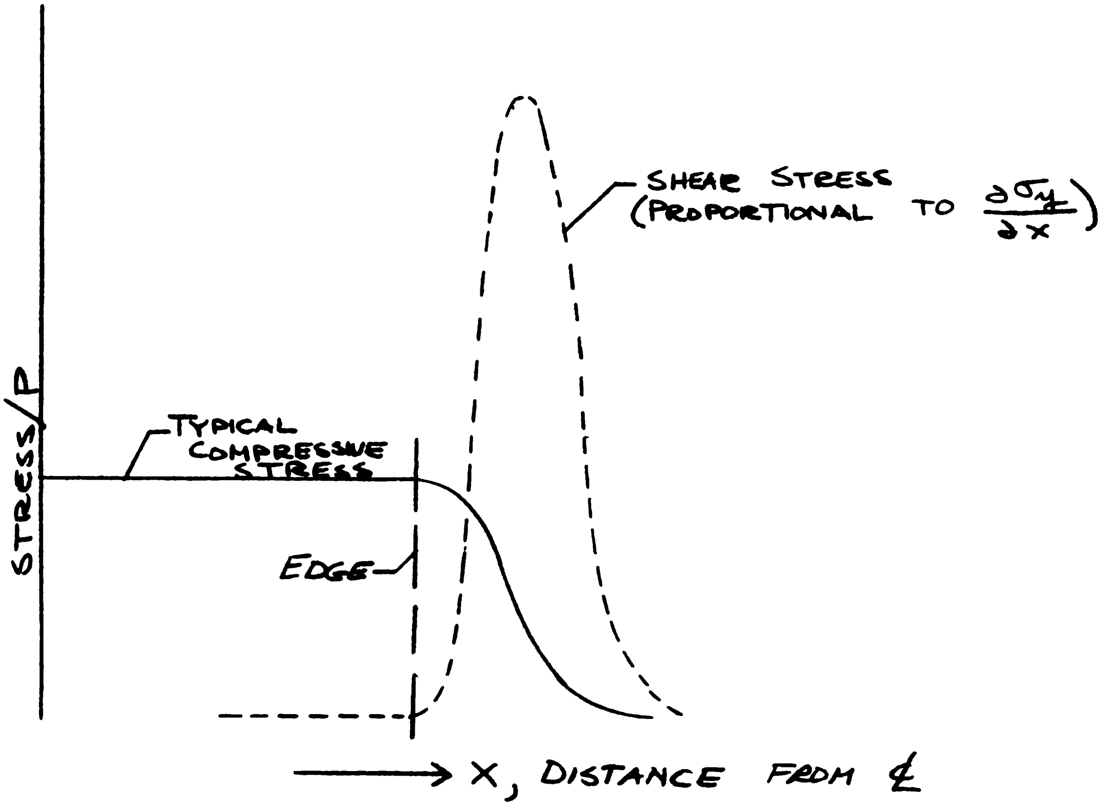


Fig. 12. Shear stress

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RESULTS OF WORK AT TEXAS A&M UNIVERSITY

Paul H. Newell, Jr.

Dr. Newell reported on:

1. design requirements for maximal support systems for the prevention of ischemia for completely debilitated individuals;
2. the construction of such a prototype system;
3. a pressure-distribution measuring system which was designed and constructed for the purpose of preclinical evaluation of any type of support system; and
4. a total tissue systems analytical approach which includes a quantifying formulative model for active transport which, if pursued, would establish a rational basis for delineating both analytical and experimental subsystem investigations.

It was suggested that the model proposed will contribute to the understanding and, consequently, the avoidance of many clinically important phenomena such as edema, reactive hyperemia, and the destructive metabolic and catabolic effects of tissue denervation.

SUMMARY OF INVESTIGATIONAL WORK AT RANCHO LOS AMIGOS HOSPITAL

Vert Mooney

Our investigational work has been divided into three areas: search for improved materials to supply support for tissues; measurement of potentially destructive forces offered by support materials; and analysis of tissue activity in a rigid, total-contact environment.

The work with materials has led us into evaluation of the ability of various materials to compress under high-pressure loading. As an outgrowth of this, various new designs of material have developed such as cross-hatched, polyurethane foam, Plastazote (expanded polyurethane) formed over plaster positives, and Adaptifoam (resin-filled polyurethane foam). The rationale for the use of each of these materials must be explained so that expectations will not exceed the limitations of the material. Measurement of the forces involved is necessary. To do this an electropneumatic pressure transducer was designed and constructed. The cross-hatched polyurethane foam was used in an effort to provide a situation where shear forces would not be translated from one location to another, but water vapor would be allowed to escape. Plastazote can be formed at a temperature tolerated by tissues. Thus, the construction of shoes and inserts is reasonable. To avoid packing down, various maneuvers using wafering and molding have been utilized.

Finally, in an effort to avoid the excessive forces of the membrane (hammocking) of support devices, a specialized material, Adaptifoam, was developed by Mr. John Rogers, Medical Engineer on the Rancho Los Amigos Hospital staff. The rationale for its use is based on tissue contouring which will not "bottom-out" and, thus, give total, even support.

In addition to the electropneumatic pressure transducer, emphasis has been focused at the measurement of temperatures which theoretically reflect the severity of destructive forces. Based on ideas supplied initially by Dr. Paul Brand, differential temperature measurement has been made at areas where destructive forces can be localized. These support the view that the longer the temperature elevation persists the more overwhelmed body-compensatory mechanisms are and, thus, the more destructive the forces may be.

Understanding the events occurring under rigid-support devices such as plaster casts is important when we utilize these materials to treat postoperative wounds with the threat of edema and edematous, infected feet. We have been able to show that in a totally rigid, total-contact dressing edema does not form after surgical trauma; and tissue pressures tend to slowly resolve. In addition, ambulation in a total-contact cast will cause a loss of edema and decrease in volume of the limb. The total-contact cast has been very effective in treating diabetic ulcerations and postoperative amputations.

RANCHO FLOTATION BED

James B. Reswick

The Rancho flotation bed consists of a tub filled with warm oil-well-drilling mud, a thick fluid twice as heavy as water. The surface of the mud--actually a stabilized colloidal suspension of barites--is covered with an extremely loose-fitting vinyl sheet. The patient lies on a normal bed sheet loosely arranged on the vinyl membrane. Because of the looseness of this membrane, the patient literally floats in the heavy fluid with only half of his body below the surface. He is thus supported by hydrostatic pressure alone which does not exceed about 25 mm Hg (.43 psi) when he is in the supine position. All parts of his body--head, arms, torso, and legs--are held in a natural, straight-line position without any points of high pressure. Wound dressings, braces, or casts merely depress into the mud without causing local pressure or friction.

The bed is intended for use with paralyzed patients who may be susceptible to pressure sores; patients for whom the hard surface of a normal bed would produce discomfort; and for patients where friction on a wound might retard healing.

The Rancho flotation bed was designed to achieve the following objectives:

1. To provide a bed in which the maximum pressure on the body is well below that which might cause a pressure sore.
2. To support the patient in a comfortable, straight-line position to prevent flexor contractions of the hips and knees.
3. To provide a bed in which patients wearing braces or casts may be more comfortable.
4. To provide a bed in which temperature is controlled for comfort and control of skin temperature.
5. To provide a bed in which the patient lies substantially above the surface of the bed so that he does not feel confined, and so that catheters and other devices may function normally.
6. To provide a bed which is relatively inexpensive and "fail-safe," and in which the patient may be easily managed by nurses and therapists.

PRINCIPLE OF DESIGN

It is well known that a body when floating in water is supported by hydrostatic pressure alone. Such pressure acts perpendicular to the surface of the body and is incapable of exerting shear or friction tangential to the surface of the body. Hydrostatic pressure increases with depth of emersion and in the case of water reaches 25 mm Hg (.43 psi) at a one-foot depth. The density of the human body is only slightly greater than that of water. This is proven by the swimmer who can easily float when his lungs are inflated, but who sinks when he exhausts his air. A person floats readily in salt water which is denser than fresh water and in the case of very salty water, such as is found in the Dead Sea, a human will float with a portion of his body above the water.

If one continues the analogy to consider a human being floating in a fluid twice as heavy as water, it is easy to visualize that he will displace a volume of fluid equal to about half that of his body and that he will float with half of his body above the surface. Under these conditions, it is interesting to note that the pressure situation is the same as if he were floating just under the surface in water since he sinks only half as far into a fluid which is twice as heavy. Thus the maximum pressure acting on the deepest part of his body which, if assumed to be six inches, is 25 mm Hg (.43 psi). At most other parts of his body it is very much less. Since the human body is roughly symmetrical about its center line, the back and other parts of the body will be kept substantially straight whether the patient is lying prone, on his side, or supine.

It is vital that the sheet which separates the patient from the fluid be very loose and incapable of exerting any tension. This results in a highly wrinkled surface which at first sight might appear to cause discomfort or local pressure under the folds. Actually, such wrinkles easily dip into the fluid and the patient is unaware of them. The cover is arranged with a considerable excess of material so that it is easy to turn the patient since he is floating with his center of gravity at just about the axis of rotation.

In order to provide a firmer surface for dressing the patient and other nursing functions, an air mattress is built into the vinyl sheet so

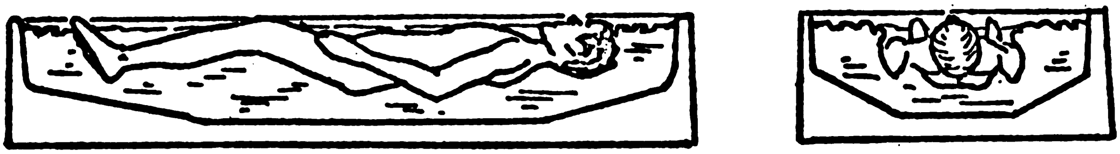

Fig. 1. Patient in water flotation bed is virtually submerged with marginal stability.

Fig. 2. Patient in Rancho flotation bed (mudbed) is stably supported well above surface of bed.

Fig. 3. Air mattress provides firmer surface for management.

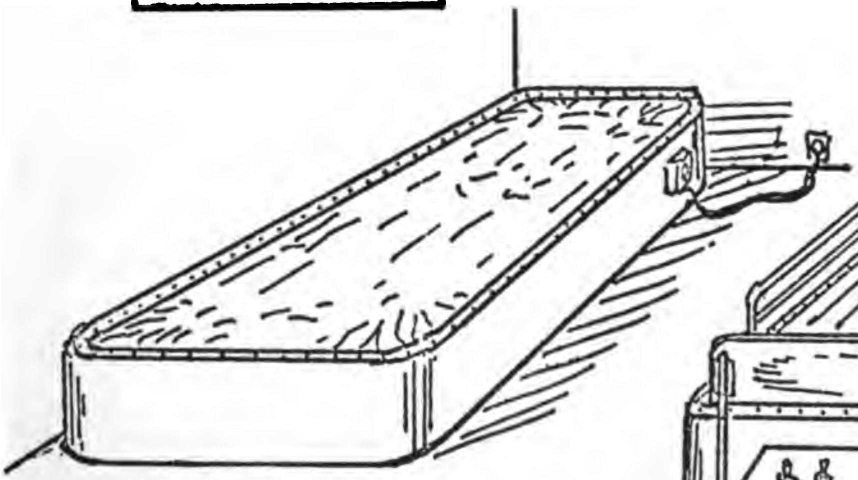

Fig. 4. "Home-type" version is simply a tub with heated mud and vinyl sheet.

Fig. 5. Hospital-type Rancho flotation bed.

Fig. 6. Person on home-type water bed is supported more by hammocking of cover than by hydrostatic pressure. (JBR 15 Oct 1971)

that, when inflated, it tends to float the patient to the surface and provide a firmer space for handling the patient.

The mud is heated electrically to about 85 deg. F. so as to be comfortable and to control effectively the surface temperature of the patient and thus regulate uncontrolled sweating. The large mass of mud heats up and cools down very slowly--a matter of days--so that the patient is protected from sudden temperature changes in case of maladjustment of controls or power failure.

About 50 gallons of mud, weighing about 800 pounds, are used in a somewhat sculptured tub to minimize the overall weight, but still to provide support for the patient in various positions.

EXPERIENCE TO DATE

A single prototype bed has been evaluated in the Spinal Injury Service at Rancho Los Amigos Hospital. A high-level-spinal-injured male has used the bed for over five weeks. During this time nurses have reported no major management problems. Pressure sores on his back and elbows, which he presented at the time he was put into the bed, have healed. Pressure sores on his heels, which he also presented, have improved. The patient has worn a head brace with supporting structure on his chest and back with no discomfort. During this period of five weeks, he was never turned at night while sleeping. No reddened areas or other evidence of pressure occurred. The patient was subjected to range-of-motion exercises and commenced a wheelchair conditioning program during the day.

DISCUSSION

The Rancho flotation bed, a relatively inexpensive bed, operating on a new principle for the prevention and therapy of pressure sores, has been built and initially evaluated at Rancho Los Amigos Hospital. Initial experience with one patient over a five-week period indicates that all of the design objectives have been achieved. It should be emphasized that the bed was designed for the primary purpose of avoiding pressure sores without the need to turn the patient. There are many well-known adverse effects related to bed rest, including potential blood pooling, respiratory distress, contractures, and lowering of motivation of the patient. A bed such as the Rancho flotation bed requires a carefully designed nursing and therapy program to supplant the lack of body movement which it permits from a pressure-sore-prevention point of view.

POINT MEASUREMENT OF NORMAL PRESSURES

James L. Cockrell

This report concerns point-measurement pressures perpendicular to the surface between anatomic tissue and a mechanical device. The equipment, methods, and results described were developed for application to forces encountered in weight-bearing limb prostheses. In general these pressures are much higher than systolic blood pressure, and are primarily cyclic in that they occur in excursions to a maximum and sometime in the cycle return essentially to zero. A discussion of transducers, digital data storage, and display is given. Sample data for a small study comparing liner materials in patellar-tendon-bearing prostheses are presented.

Fig. 1. Pressure-transducer arrays on below-knee stump.

The smallest area over which pressure is monitored is that provided by the Kulite semiconductor strain-gauge pressure transducer which is approximately 1/8 in. in diameter and .030 in. thick. Four units are arranged in a square pattern about 1 in. on a side with a fifth transducer in the center, or in a line about 2 in. long. Figure 1 shows the square array mounted at the patellar-tendon site and the in-line array mounted at the lateral tibial condyle site. Pressure data are combined with information on knee and hip angles and foot-switch data to present a complete picture of gait cycle.

The analogue data from the transducers described are converted to digital form by means of a laboratory computer and stored for use in a number of display schemes. Figure 2 represents the time history of a single transducer and shows the variations during a single cycle. Figure 3 shows all the data for each instant of time (measurements taken at 20 millisecond intervals). Instantaneous pressure data are shown on the manometers at the left, a stick figure of tibia and femur shows simulated leg positions, and a foot outline with numbers gives closures on four foot switches (two on the heel, one on the ball of the foot, and one at the toe). In addition, the test number and frame number indicating the particular position in time are also given. These and other displays are described in detail by Rae and Cockrell[1].

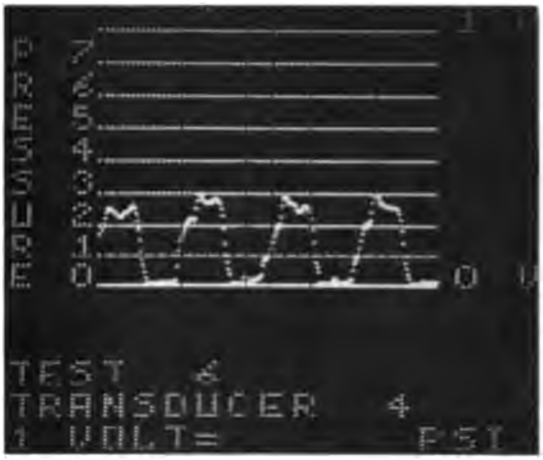

Fig. 2. Variation over four steps in patellar-tendon region (full scale = 40 psi).

Fig. 3. Leg-foot-pressure display (full scale "maximum" in 40 psi).

An example of the kind of study that can be made using point-pressure measurements is the comparison of liner materials for the PTB prosthesis. The University of Michigan Medical Center for the past few years has been experimenting with a gel liner. Specific comparison is made with a hard socket and with the conventional Kemblo liner[2]. The principal results are shown in Figure 4. Because the instantaneous pressures during a walking cycle vary over a wide range, the comparison is made with the average of the

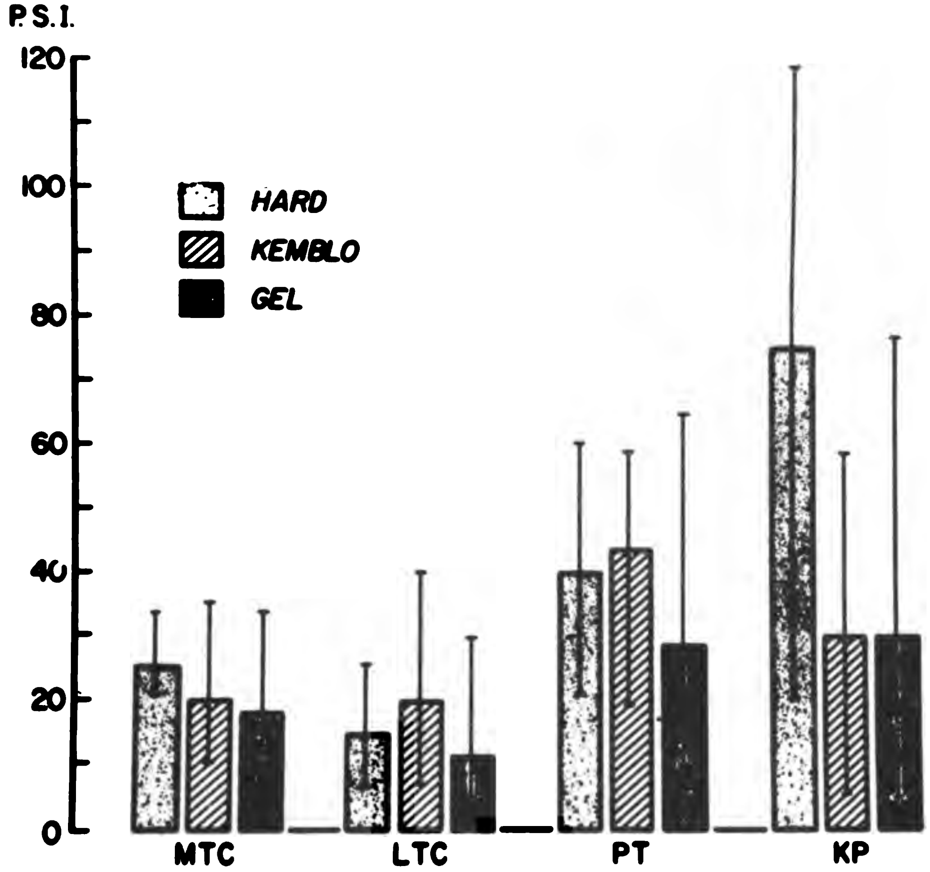

Fig. 4. Comparison of average peak pressures at four critical BK sites--for three types of socket liners.

peak pressures that occur during a five-second test. Note that in this particular situation the average peak pressures were somewhat lower for the gel liner at each of the four sites studied. These sites were the patellar-tendon region, the anterior-distal segment of the tibia and the lateral and medial condyles of the tibia.

A significant conclusion of the study was that the small difference seen between the average peak pressures for the various liner materials does not account for the very significant subjective observations and experience of patients using these liners. The patient acceptance of the gel liner has been so conclusive in this Center that the pressure parameters observed become somewhat suspect. Consequently it is felt that either the appropriate measure of normal pressure is not being tabulated or more probably the missing shear component is the significant variable.

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SENSORY AUGMENTATION IN WEIGHT-BEARING TRAINING

A. Willem Monster

There are a number of physical disabilities in which patients experience difficulty in properly loading the lower extremity during locomotion or under various semistatic postures. Examples of these are lower-extremity amputees, patients with hip fractures, and patients with sensory loss due to either peripheral (such as neuropathies in diabetics, uremics, alcoholics, leprosy, etc.) or central neurological disorders. Because these patients have lost some or all of the normal cues to effect well-controlled actions, they are not sure how to properly distribute and support their weight in order to retain their balance and thus prevent falling.

Objective training devices which provide the patient with a "sense of performance" can in many cases reduce the incidence of reinjury, or insult, and increase the quality of various functional modes of operation of the patient. The specifications for such sensory assistive devices are disability-related to a great degree, and feasibility studies on Sensory Compensation in general are currently in a very preliminary stage. Many questions concerning 1) the limitations of sensory modalities to utilize information, 2) the interaction between modalities, and 3) the relation between peripheral and central components (memory, sense of innervation) in the execution of various sensorimotor activities, are currently unanswered.

The organization of the neuromotor system permits a functional integration of many contributions, while the existence of feedback loops allows an efficient and reliable exchange of information between central and peripheral subsystems. Several of these closed loops have been identified at different levels of the neuraxis (Fig. 1), but controversy as to their functional significance exists. When one or more pathways or structures are permanently damaged so as to preclude adequate "internal" compensation, a functional disability results. In an effort to improve the treatment of such a disability we have directed our attention to the measurement of load

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This work is supported under RT-8 Grant Project No. R123 and Research and Development Grant No. 2936M from the Social and Rehabilitation Service.

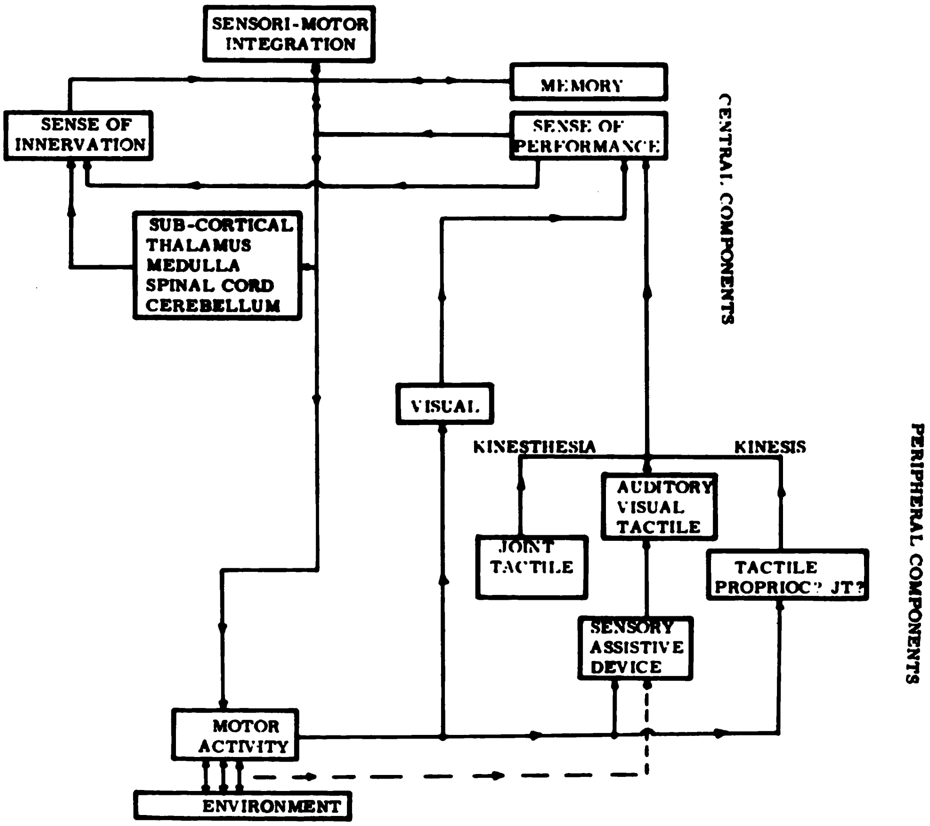

Fig. 1. Relationships between components contributing to functional motor behavior.

and load distribution in the insensitive limb for the purpose of sensory augmentation in weight-bearing training.

DEVELOPMENT OF DEVICES

At the inception of this project the most suitable force transducer contemplated was one having the form of a disposable, thin, flexible insole for a shoe. On the basis of experience with the ability of normal subjects to judge limb loading, the required accuracy of such a transducer was arbitrarily specified to be ± 10 per cent of the static loading; i.e., all weight on one foot, but irrespective of its distribution over the weight-bearing surfaces of the foot.

A survey of recent approaches to force instrumentation was made to determine the feasibility of the contemplated configuration. Most researchers have directed their attention toward reducing rigid transducer dimensions to a size sufficiently small in comparison to the radii of curvature of pressure-bearing surfaces, so that flexural forces become negligible[2]. This approach is, however, generally quite expensive to implement.

Two other approaches to flexible transducer design are potentially much less expensive--pressure-variable resistance and pressure-variable capacitance. The first approach makes use of the change with applied pressure of bulk resistivity of an elastomeric material which has particles of conducting material imbedded throughout. A major problem encountered is the inability to achieve uniform particle distribution within the elastomer. To function satisfactorily in this application the transducer must respond uniformly to total applied force independent of the pressure distribution. Because such response is dependent on the uniformity of the material, the resistance approach, although potentially the least expensive, does not appear feasible at this time.

A somewhat flexible variable capacitance transducer, the FILPIP*, was developed in the fifties at the Franklin Institute. However, researchers have reported a number of problems with the device[1]. Moreover, because of its small (3/4 in.) diameter active area, an expensive array would be required for our purposes.

Static testing of foam tapes available from Minnesota Mining and Manufacturing Co. showed that such a material might be suitable for use in a transducer which would have the characteristics desired. A testing apparatus was then designed and fabricated in order to establish the dynamic properties of these materials. The device is a closed-loop servomechanism consisting of an electronic platform scale, servoamplifier, solenoid, mechanical linkage, and power supply (Fig. 2). The scale consists of an unbonded strain gauge load cell and amplifier. The output of the scale and an externally supplied command signal are compared at the input of the servoamplifier; the amplified difference signal drives the solenoid causing force to be applied through the mechanical linkage to a pressure plate which bears upon the transducer sample on the scale platform. With this apparatus, precisely controlled force functions of up to 250 pounds can be applied

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* Franklin Institute Laboratories Pressure-Indicating Patch

Fig. 2. Servo-controlled platform for testing of static and dynamic properties of elastomers.

repeatedly to samples to permit accurate comparison of different materials or different areas of the same material.

Of the tapes tested thus far, 3M #4416 Scotch Mount Tape has been found to be the most suitable for the application. Response of a test transducer consisting of three 2-in.-square plates with the #4416 tape may be seen in Figure 3. Because this tape is manufactured with an adhesive surface, transducer assembly is greatly simplified. Both the tape and the plates (currently .001 in. brass shim stock) can be cut to standard insole dimensions using scissors or inexpensive stamping dies.

The configuration currently used consists of three plates with the outer two at ground potential. For maximum sensitivity the inner plate can be tailored individually using the patient's footprint as a pattern so that only the weight-bearing surface of the foot is monitored. Since the cost of the materials is less than $1.00, excluding connecting cable which can be reused, the transducer can be considered a disposable item. It should also be noted that the transducer has numerous other possible applications beyond the scope of this project, such as differential weight-bearing using a segmented inner plate.

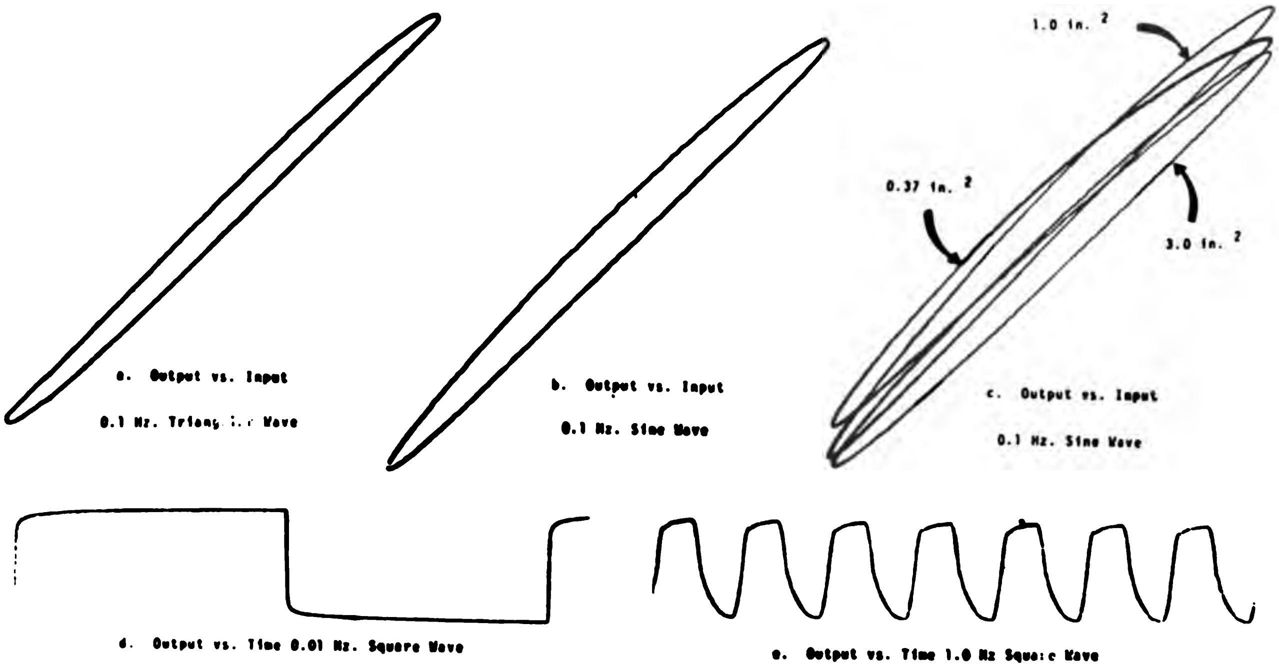


Fig. 3. Response of 3M #4416 scotch mount tape to various loading conditions.

Concurrent with transducer development was the development of circuitry to convert the changes in transducer capacitance to an electrical analogue that faithfully reproduces the forces applied to the transducer. The change in capacitance is directly proportional to the applied force of deformation in a parallel plate capacitor having an elastic dielectric which deforms in accordance with Hooke's law. If the capacitor is charged by a constant current to some fixed arbitrary voltage, and then rapidly discharged to zero in a relaxation oscillator, the period of oscillation is inversely proportional to the charging current. Thus, if the charging current is maintained so as to keep the period of oscillation constant with changing capacitance, the charging current will be directly proportional to capacitance.

In the present circuit configuration, the transducer oscillator and a similar fixed period oscillator gate each other on and off alternately through a flip-flop. This permits comparison of the period of the transducer oscillator with that of the fixed oscillator because the duty cycle of the flip-flop output will reflect any change in the period of the transducer oscillator. The flip-flop outputs are connected to the inputs of a differential amplifier through low pass filters such that any deviation from a 50 per cent duty cycle will produce an error voltage. The amplified error voltage controls the charging current of the transducer oscillator so as to maintain the 50-per-cent duty cycle, and also provides the required electrical analogue proportional to applied force.

The 3M #4416 material has a negative .005/C° thermal expansion coefficient. Thus greater charging current must be supplied with increasing temperature to keep the oscillator period constant. When required this may be compensated by monitoring the ambient temperature at the transducer end of the connecting cable with a thermistor, and using the thermistor to control a shunt across the transducer such that proportionally less current flows through the shunt with increasing temperature. Because the insole is in close contact with the sole of the foot, the temperature is quite stable and therefore temperature compensation will normally not be necessary.

Because the outer plates of the transducer are at ground potential they act as an electrostatic shield for the inner plate. The unstressed capacitance of the transducer is approximately 15 pf/in^2; the capacitance of miniature shielded cable (50 pf/ft) is of the same order as that of a typical transducer. In order to provide shielding for the conductor to the inner plate of the transducer without introducing the effects of stray capacitance into the system, a boot-strapping technique can be employed whereby the shield is connected to the low-impedance output of a high-impedance input, noninverting, unity gain amplifier whose input is the inner conductor. Because the shield tracks the inner conductor it sees no capacitance[3].

A sizable portion of instrumentation development was concerned with effective display of the processed transducer output. The simplest display is direct meter reading. However, it is felt that an auditory display would be less likely to distract the subject from visual balance cues. One of the simplest yet highly effective means of accomplishing a proportional auditory display is by frequency control of an auditory signal such that it passes through zero Hertz at the desired loading level. This proportional feedback signal has so far been implemented in all prototypes used for training purposes.

A more complicated, but less obtrusive, nonproportional auditory display system has been developed whereby the subject is provided tones only to indicate weight-bearing errors. If the upper limit is exceeded, a high tone is sounded; if the lower limit is not exceeded during loading, a low tone is sounded as the limb is unloaded. Figure 4 shows a transducer connected to a complete battery-powered display system. Because most components are low-power MOS integrated circuits which are exceptionally insensitive to battery voltage variation (RCA CMOS series #4000A), the effective life of the battery is approximately 50 hours.

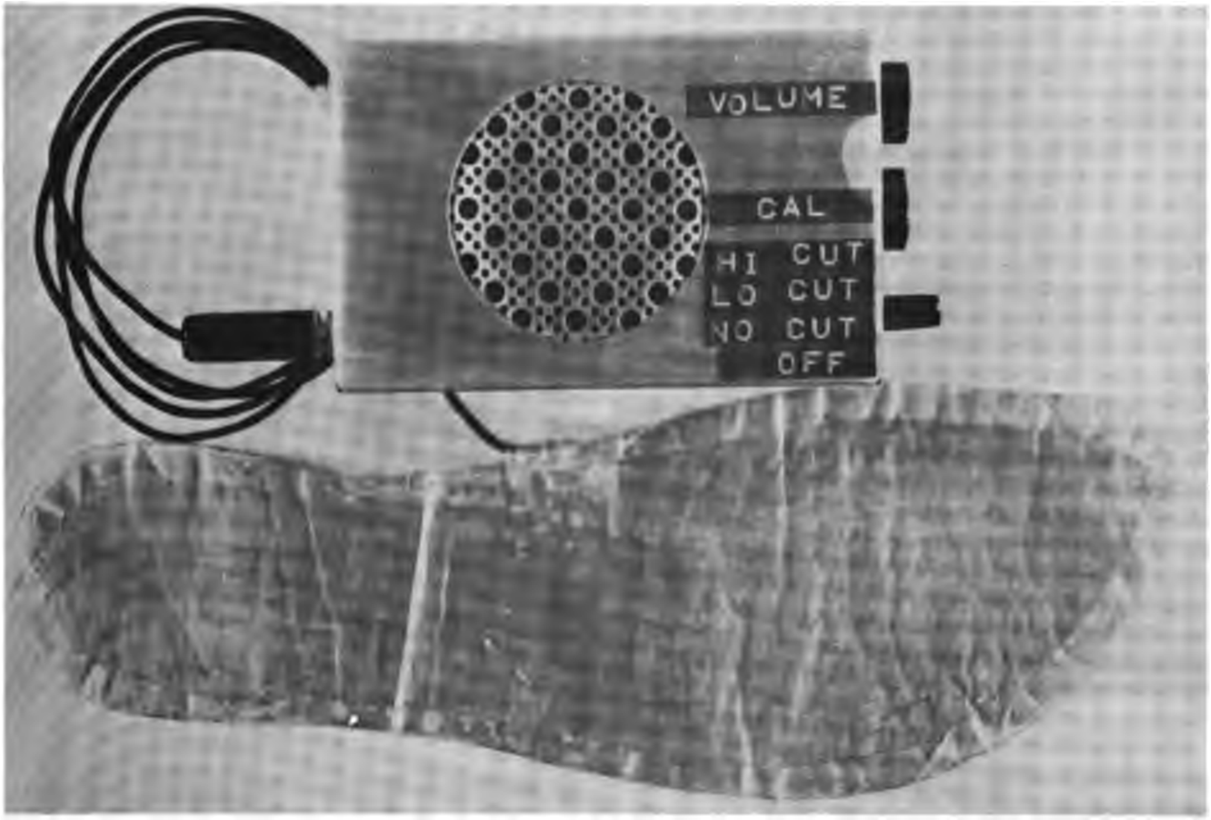
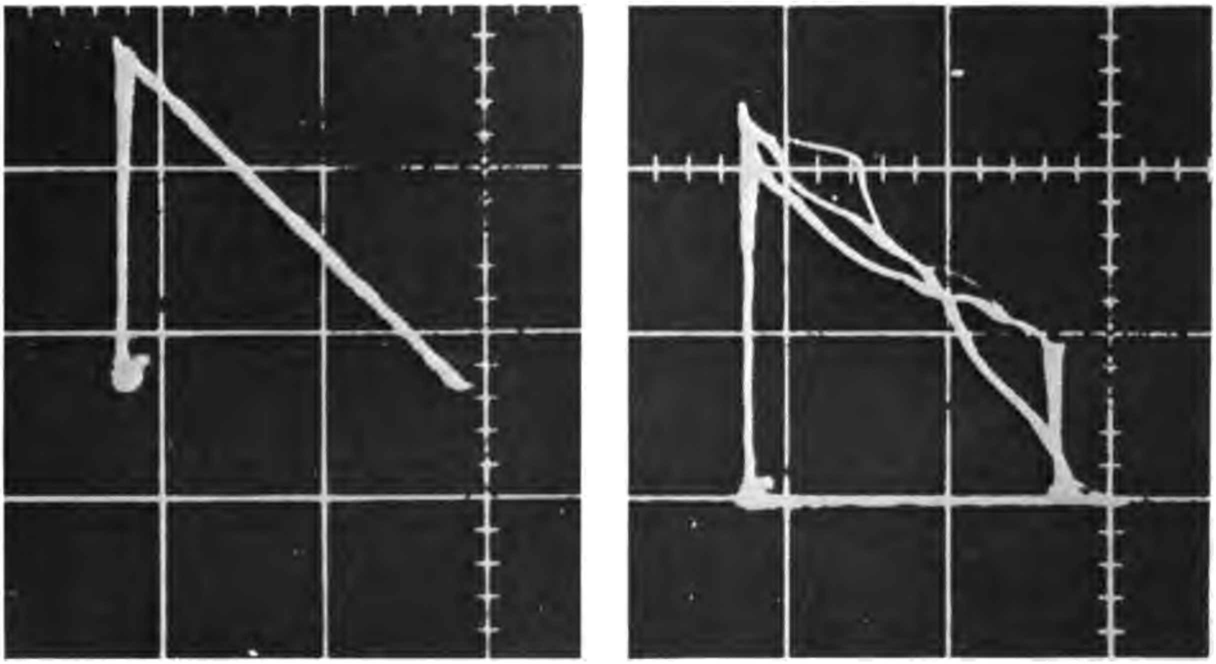


Fig. 4. Prototype of the Krusen weight-bearing monitor.

TEST RESULTS

In the course of development, a number of preliminary clinical tests were made to evaluate the prototype. Two parallel force plates (each 5 feet long) and transducing and monitoring equipment were used. The first tests consisted of a measurement of the static (e.g., body weight) and dynamic (e.g., deceleration mass) components in the vertical weight-bearing forces during various speeds of walking. Under a semistatic condition in which the feet are lifted and placed very slowly, the sum of the forces on the two plates is a constant and equal to the body weight. When the force on one leg is displayed horizontally and that on the other vertically, the result is a straight line (Fig. 5a). During normal locomotion the dynamic component plays a significant role (Fig. 5b) and must be taken into account when calibrating the reference load of the monitor. During a maintained posture, a source of variation in the applied forces is caused by body sway.



a b

Fig. 5 (a and b). Effect of various walking speeds on size of static and dynamic component of the total load. Under primarily static conditions the sum of the load on right (vertical axis) and left foot (horizontal axis) is a constant (5a); during faster walking speeds the dynamic (deceleration/acceleration) force causes a deflection from the straight line (5b).

Figure 6 shows this variation for a given semistatic weight-bearing distribution (60 per cent of total weight on left and 40 per cent on right limb), maintained for approximately 30 seconds; a peak-to-peak variation was observed which is about 10 per cent of the subject's weight. These measurements provided some indication for the required range and static resolution

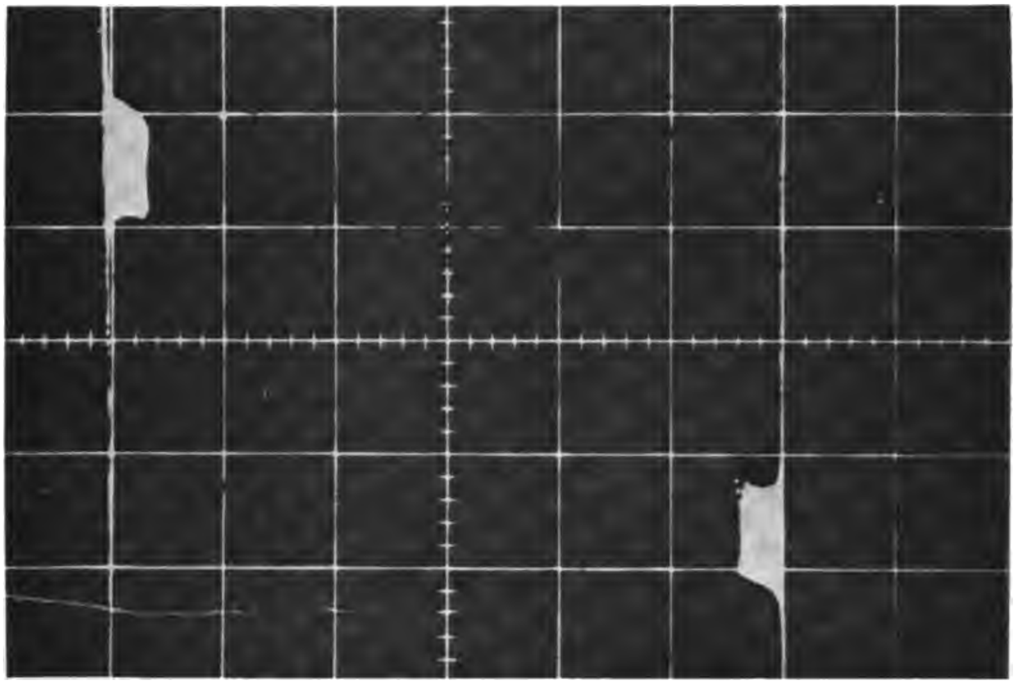


Fig. 6. Variations in the amount of load on right and left foot due to postural swing. The maximum variation is represented by the size of each of the two vertical bands as they appear on the memory oscilloscope. The vertical distance between the center of each band corresponds to approximately 20 per cent of body weight (60 per cent on left foot, 40 per cent on right foot).

of the transducer. The next step was to consider the difference in dynamic characteristics between the force plates (natural frequencies > 50 c/s) and the transducer, placed as an insole inside the subject's shoe.

A normal walking pattern (left foot only) is shown in Figure 7a, in which the essential difference in transducer bandwidth is clearly indicated by the peaks in the force registered by the force plate (first cycle) and the smooth curve of the transducer output, on the two superimposed curves. This is further illustrated in Figure 7b which shows a difference in the forces registered by the transducer (Y coordinate) and the force plate (X coordinate due to the dynamic (phase) lag in loading and unloading cycle). The device was then fitted to patients undergoing weight-bearing training, e.g., amputees. The fitting procedure is extremely simple because

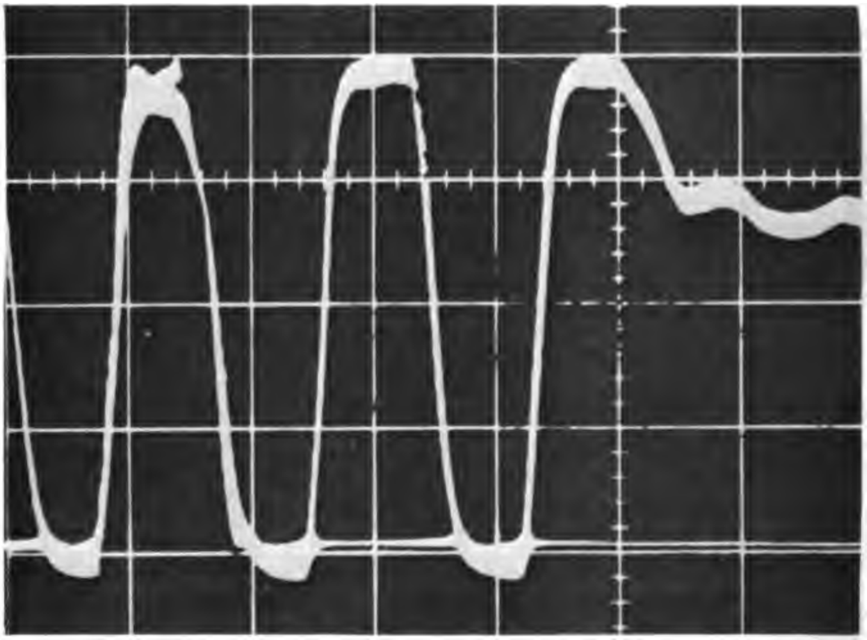


Fig. 7a. Illustration of the difference in band-width between walking platform and insole transducer as shown in a load vs. time plot during normal walking speed, with two superimposed traces. The sharp peaks in the first cycle are the platform response.

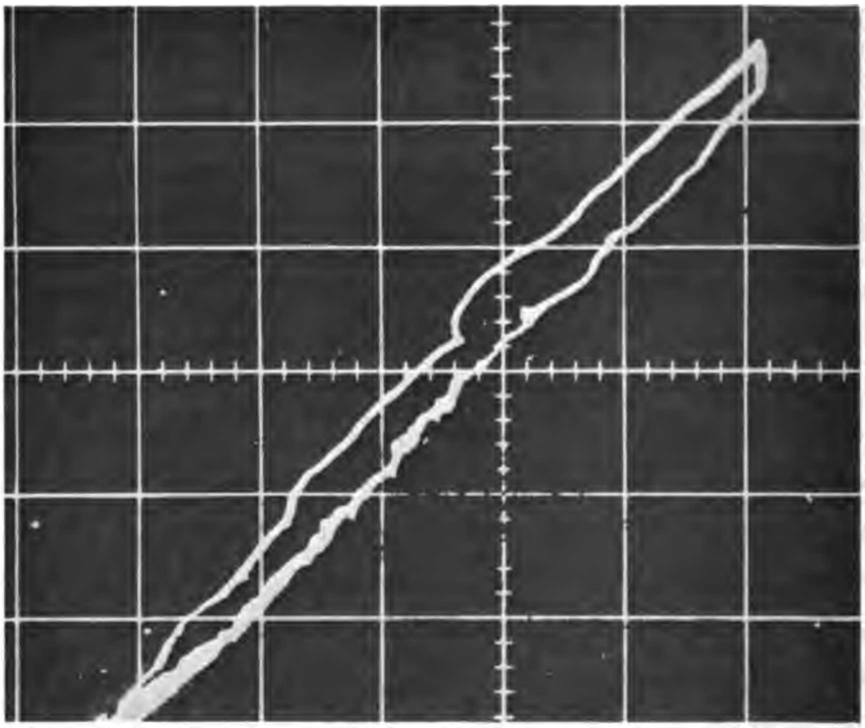


Fig. 7b. Dynamic lag between walking platform and insole transducer during fast loading/unloading cycle.

the transducer itself can be cut to fit any shape (shoe or foot) and then calibrated by placing the subject on a scale. The load-bearing force and desired range are calibrated using the same scale, and the subject is given some practice time before tightening up the range to the desired minimum.

The patient's performance during the practice was monitored with the force plates (Fig. 8), to spot the source of any difficulties in the patient's ability to apply the desired percentage weight-bearing on each limb. Training was continued until a range of variation in the specific weight-bearing load of 10 per cent body weight could be maintained reliably during a comfortable walking speed.

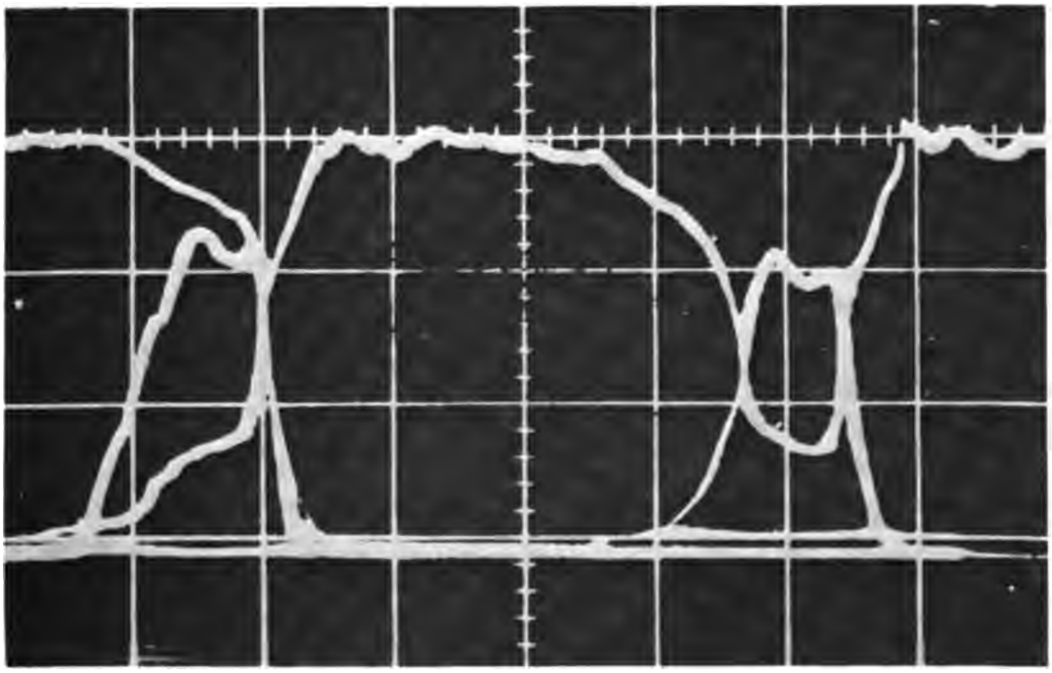


Fig. 8. Partial weight-bearing training (the load on both right and left foot is shown) with maximum load for left foot set at 40 per cent of total weight. Performance was monitored with walking platform and auditory display from insole transducer.

There are many other therapeutic applications for the transducer, such as training for balance, foot placement, etc., by building the inner plate of the insole out of sections. Reduction in size is possible to a few tenths of an inch without interfering with the resolution. Because a reliable load-monitoring package can be made available for very reasonable cost, emphasis is now being placed on the production and clinical evaluation of the device.

Other applications of the material might involve the measurement of forces on supporting surfaces of the body. Because the material is quite thin and can be shaped to accommodate curved surfaces with large radii, it could possibly be used also for determining local socket pressures in limb prostheses and other orthopedic applications.

SUMMARY

A versatile and inexpensive pressure monitor is presented. Both the transducer itself, a closed-cell elastomer that is commercially available, and the audio feedback device have numerous applications, of which one is described in some detail.

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3. Maine, R. E., and D. W. Lewis, Pressure transducers for prosthetics, IEEE Transactions Industrial Electronics Control Instrumentation (USA), IECI-17:2:138-143, April 1970.

SECTION IV. SUMMARY AND RECOMMENDATIONS

For the final session the Chairman presented a matrix consisting of status of knowledge and suggested priority versus tissue systems between the bone and the external force. The status of knowledge and an estimate of priorities were discussed for each "box" in the matrix and are shown in Table 1.

Additional recommendations are:

1. Further development of instrumentation to measure the forces applied to soft tissues. Methods for use in clinical practice are required in addition to sophisticated methods for experimental purposes.

2. Further studies concerning the effects of frequency of application forces.

3. Studies for definition of the differences between pathologic and protective responses for involved tissues and systems such as skin, subcutaneous tissue, muscle, collagen, tissue fluids, and circulation.

Table 1. Status of knowledge and priorities on the effect of pressure on soft tissues

	Conditions of loading resulting in tolerated ischemia		Maximum tolerated loading		Beneficial loading (conditioning)		Study of mechanical environment for prevention of damage to denervated tissue	
Tissue system between bone and external force	Constant	Intermittent	Constant	Intermittent	Constant	Intermittent	Optimal	Tolerated
	4/A	3/A	0/A	0/A	0/B	0/B	2/A	1/A
Skin	2/A	2/A	1/A	1/A	0/A	0/A	2/A	3/A
Subcutaneous Tissue	2/A	2/A	0/A	0/A	0/A	0/A	2/A	2/A
Circulation	2/A	2/A	0/A	0/A	0/A	0/A	0/A	0/A
Fascia	0/B	0/B	0/B	0/B	0/B	0/B	0/B	0/B
Peripheral Nerve	0/C	0/C	0/C	0/C	0/C	0/C	0/C	0/C
Muscle	0/D	0/D	0/D	0/D	0/D	2/D	0/D	0/D
Scar	0/A	0/A	2/B	2/A	2/B	2/B	1/C	1/C
Tissue Fluids	0/C	0/C	0/C	0/C	1/D	1/C	0/B	0/B

Top number is state of knowledge on a 0-5 scale with 0 representing no information.
Bottom designation is priority on an A-E scale with A being top priority.

APPENDIX A

WORKSHOP ON EFFECT OF PRESSURE ON SOFT TISSUES

COMMITTEE ON PROSTHETICS RESEARCH AND DEVELOPMENT
NATIONAL ACADEMY OF SCIENCES--NATIONAL RESEARCH COUNCIL

Texas Institute for Rehabilitation and Research
1333 Moursund Avenue
Houston, Texas

September 21-22, 1971

9:00 am September 21

I. STATEMENT OF OBJECTIVES — Donald B. Kettelkamp, Chairman

A. Problems in Management of Amputees and Paralytics — Vert Mooney

B. Problems in Management of Insensitive Limbs and Extremities — Paul W. Brand

C. Problems in Management of Other Neurological Diseases — Richard M. Herman

II. BASIC BIOLOGIC CHARACTERISTICS OF TISSUES AND SYSTEMS

A. Histochemistry of Skin — Jeffrey Pinto

B. Microcirculation — Mary Wiedeman

C. Mechanics of Edema — Fred Caldwell

D. Effect of Stress on Collagen — John Madden

III. SUMMARY OF WORK CARRIED OUT TO DATE

A. Informal Presentations

1. James B. Reswick and O. Lindan
2. Paul W. Brand
3. Leon Bennett
4. Edward Peizer
5. Paul H. Newell
6. Vert Mooney

B. Discussion

IV. PROBLEMS OF MEASUREMENT

A. Informal Presentations

1. Edward Peizer
2. James B. Reswick
3. James L. Cockrell
4. W. Monster
5. Paul W. Brand
6. Others

B. Discussion

V. SUMMARY AND RECOMMENDATIONS

WORKSHOP ON THE EFFECT OF PRESSURE ON SOFT TISSUES

PARTICIPANTS

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